MEC '11

Raising the Standard

University of New Brunswick's
International Conference on
Advanced Limb Prosthetics

August 14 – 19, 2011
Fredericton, New Brunswick, Canada

Symposium Proceedings

Institute of Biomedical Engineering
University of New Brunswick

UNB

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ISBN: 978-1-55131-160-9

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Welcome to MEC '11

To all MEC'11 participants:

On behalf of the members of the Institute of Biomedical Engineering I am pleased to welcome you to the 2011 Myoelectric Controls Symposium, MEC'11. The MEC symposium has been offered since 1972, later becoming a triennial symposium, and is geared toward professionals involved in both the research and application of myoelectrically controlled limb prosthetics. Several courses and clinics are offered; research, development and application papers are presented; state-of-the–art technologies are showcased by manufactures and vendors; and opportunities provided for interaction between the various disciplines.

Given the ever-increasing function of control systems and prosthetic devices, the theme of MEC'11 is **Raising the Standard** in prosthetics training, fitting, and assessment. Towards this end the Scientific Committee has put together an exciting program of keynote speakers, research and clinical presentations, and manufacturers' demonstrations. The three keynotes are internationally recognized leaders in the rehabilitation field, providing insights in many aspects of limb prostheses development and application. There are over 85 presentations, contributed by participants from around the world, and covering important aspects of fitting, training and assessment. Representatives from all the major manufacturers of limb prostheses are exhibiting their latest developments in myoelectrically controlled devices. The scientific program activities take place at the university's Wu Conference Center.

MEC'11 has received financial support from a number of sponsors, and we thank the manufacturers, and provincial, regional and federal organizations for their continued participation.

In keeping with traditional Maritime hospitality a number of social activities will allow time for relaxation and networking with friends and colleagues. A wine and cheese reception will be held at Fredericton's new Convention Centre, and a banquet dinner at the Student Union Building's Atrium.

We hope you will find the scientific program stimulating, and that you enjoy the lovely setting of UNB and Fredericton.

Philip A. Parker
Organizing Committee Chair

MEC `11 ORGANIZING COMMITTEE

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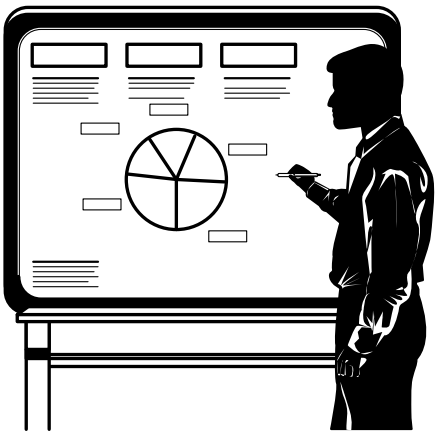
Poster Sessions

Due to the number of posters to be presented we will be dividing the presentations into two sessions: **Session A will be held on Wednesday, August 17** and **Session B on Thursday, August 18**.



On Wednesday, during the morning break and lunch, delegates will be able to view **Session A** posters. During the afternoon break, presenters will be at their posters, available to answer questions.

On Thursday, during the morning break and lunch, delegates will be able to view **Session B** posters. During the afternoon break, presenters will be at their posters, available to answer questions.

Social Events

Welcome Wine & Cheese Reception

On Wednesday, August 17, a Wine & Cheese reception, sponsored by Otto Bock Healthcare, will be held at Fredericton's new Convention Centre. The Convention Centre is conveniently located on Queen Street, across from the Crowne Plaza Lord Beaverbrook Hotel.

Banquet Dinner & Dance

On Thursday, August 18, a Banquet Dinner & Dance will be held at the University's SUB Atrium. Music will be provided by the local band "Southern Drive ", featuring drummer Jody Vincent. Jody has been a client at our Clinic for many years.

The buffet style meal will include seafood chowder, lobster, salmon, roast beef, and all the fixings.

Group Photo

Keeping with tradition, we will have a group photo taken at the Wu Conference Centre on Friday morning, during the Refreshment Break.

Wireless Internet Access

Inside the padfolio, each delegate will find instructions for your wireless access account, while attending MEC ’11. The accounts will be valid for the week of MEC ’11.

Notice Regarding Audio/Video Recording and Photography of Events

University of New Brunswick Institute of Biomedical Engineering (UNB IBME) may elect to take photographs of people and events during the MEC’11 Workshops & Symposium, from August 14 to 19, 2011. By attending MEC’11, you agree to permit UNB IBME to use your likeness in these photos in promotion of the conference. The release checked off when registering indicated that you agree that UNB IBME shall be the copyright owner of the photographs and may use and publish these photographs. UNB IBME is released from any and all claims and causes of action that you may have now or in the future based upon or in connection with photographs and UNB IBME’s use of the photographs in any manner. All rights granted to UNB IBME by you in the Release are irrevocable and perpetual. You waive all rights to any equitable relief in connection with the Release and the subject matter of the Release.

Education Credits

For each morning and afternoon session, a sign-up sheet will be at the Registration Desk. A *Certificate of Attendance* from IBME will be mailed to delegates in the fall.

Clinical Prosthetic Program – 30th Anniversary

The Institute of Biomedical Engineering was founded in 1965, as the Bio-engineering Institute. The Bio-engineering group was researching 'myoelectric control'. The group was partially funded under the Prosthetics Research and Training Units" (PRTUs) funding from the Government of Canada. Four PRTUs were initiated in Canada in response to the birth defects caused by Thalidomide.

The first myoelectric fitting in Canada was in 1965 – a collaborative effort between the UNB group and a group in Toronto (at the time called Ontario Centre for Crippled Children). The UNB team developed and built the electronic hardware, while the terminal device and prosthetic fittings were done in Toronto.

The Institute of Biomedical Engineering's Clinical Prosthetics Program was established in 1981. A prosthetist, an occupational therapist and a technician were hired to provide personalized diagnostic assessments, develop a fitting plan and set the training agenda.

The team approach allows the clinic to provide fully integrated upper limb prosthetic services, with training and technical support from one location. The team follows the client through the whole process.

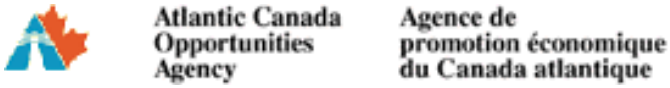
Our affiliation with the Stan Cassidy Centre for Rehabilitation allows our clients to access other clinical services such as physiotherapy, psychology, social work, adapted driving program, augmentative communications program, and orthopedic and plastic surgery.

As part of the Institute of Biomedical Engineering the clinic team is able to access electronic and engineering support when needed, and access the motion analysis lab for research purposes. The clients are able to be involved in various research projects and can be fit with highly custom devices that would otherwise not be available.

2011 marks the 30th year since the opening of the Clinic. We are happy to celebrate this milestone with you at MEC '11.

FINANCIAL SUPPORT

The Institute of Biomedical Engineering and the *MEC' 11* Organizing Committee would like to recognize the following organizations for their contributions to the symposium:



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Thank you for making this week a success.

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Wednesday, 17 August 2011

Wu Conference Centre Auditorium

8:15 AM	**WELCOME**
8:30 AM	**KEYNOTE - HELENA BURGER**

Presentation of Papers

Time	Paper Title	Presenter
9:30 AM	Assessment of Capacity for Myoelectric Control - Evaluation of Task Difficulty, Newly Merged Items and Redefined Rating Scale	Helen Lindner
9:45 AM	A Quantitative Operational Performance Measuring System for a Myoelectric Hand: A Preliminary Study	Isamu Kajitani
10:00 AM	***REFRESHMENT BREAK / EXHIBITOR SHOWCASE***	
10:30 AM	Challenges and Solutions in Control Schemes for Electrically Powered Articulating Digits	MacJulian Lang
10:45 AM	Comparison of Two Myoelectric Multiarticulating Prosthetic Hands	Brian Waryck
11:00 AM	Design of a Hydraulic Hand Prosthesis, with Articulating Fingers	Gerwin Smit
11:15 AM	Case Study: Multiple-Limb Amputees Fit with Powered Partial Hand Prostheses	Melanie S. Harris
11:30 AM	High-Fidelity Interface and the Principle of Compression-Release Stabilization	Randall Alley
11:45 AM	Controlling Two Independent Joing Motions with the Acromion	T. Walley Williams
12:00 PM	***LUNCH BREAK***	

Wednesday, 17 August 2011

Presentation of Papers (Continued)

Time	Paper Title	Presenter
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1:30 PM	Case Study: Surgical, Prosthetic, and Therapeutic Considerations for a Patient with Ipsilateral Brachial Plexus Injury and Transradial Amputation	Robert Dodson
1:45 PM	Osseointegration of the Upper Limb – Prosthetic Treatment	Stewe Jönsson
2:00 PM	Application of Haptic Feedback for Improved Prosthetic Control	Pravin Chaubey
2:15 PM	Fitting & Suspension Techniques for a Transhumeral Amputee with Burn Injuries: A Four Year Retrospective Case Study	Ryan Spill
2:30 PM	The Prosthetic Habilitation of a Congenital, Transradial Limb Deficient Child: A Case Study Analyzing the Functional Effectiveness and the Benefits of Early Prosthetic Fitting, Appropriate Prosthetic Equipment, and Consistent Caregiver Follow up.	Jennifer Peterson
2:45 PM	Occupational Therapy: Training Postural Control for Functional Upper Lim Prosthesis Use	Tiffany Ryan
3:00 PM	***REFRESHMENT BREAK / POSTER EXHIBITION***	
3:30 PM	Prosthesis-Guided Training for Practical Use of Pattern Recognition Prostheses	Blair Lock
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Wednesday, 17 August 2011

Presentation of Papers (Continued)

4:15 PM	Feedback in Voluntary Closing Arm Prostheses	Dick H. Plettenburg
Time	Paper Title	Presenter
4:30 PM	Continuous Position and Force Control of a Multigrasp Myoelectric Transradial Prosthesis	Skyler Dalley
4:45 PM	The Design of a Myoelectrically Controlled Hand for a Five-year Old Chile with Multiple Actuators	Thomas Redman
5:00 PM	A One Year Retrospective of Partial Hand Patients Using ProDigits	Diane Atkins

KEYNOTE:

PERSPECTIVE OF A PRM SPECIALIST ON REHABILITATION OF PERSONS FOLLOWING UPPER LIMB AMPUTATION

Helena Burger

University Rehabilitation Institute, Republic of Slovenia, Ljubljana, Slovenia

INTRODUCTION

The human upper limb, especially hand, is a very complex part of the body with many different functions including motor, sensory and expression. After amputation, all functions of the human hand are lost. The amputation dramatically changes a person's sense of body image, it has severe psychological consequences and it influences a person's satisfaction with life (1). Due to lost functions, a person has problems at many activities, leisure pursuits, social contacts as well as at work (2 - 9). The main aim of rehabilitation is to enable persons of any age, gender or culture to become independent in performing individual meaningful activities of daily living and to reintegrate them into society (to be able to participate in all social roles).

TEAM WORK

The key to successful rehabilitation of people following upper limb amputation is teamwork (10) which improves short- and long- term outcomes (11, 12). The team consists of the patient and his or her family, surgeons experienced in upper limb amputation, specialists of physical and rehabilitation medicine (PRM), nurses, occupational therapists (OTs), physiotherapists (PTs), certified prosthetist orthotists (CPOs), psychologists, social workers, vocational counsellors, and others, all with special knowledge and experience in rehabilitation of people following upper limb amputation. It is important that all the stakeholders are included into rehabilitation and its planning. The rehabilitation team has to contact the school for persons who are still in the educational process or the employer for those who are working and together with them find the optimal solution for the individual. Recommendation B (good practice) of British guidelines for amputee and prosthetic rehabilitation is that experienced clinical counselling and psychological support should be available to all upper limb amputees (13).

The rehabilitation team has to work on all levels of human functioning (14, 15) in an interdisciplinary way. The team also has to use valid, reliable and sensitive outcome measures to demonstrate the improvement and the effects of work. All team members have to participate in the research work. Unfortunately there is only little low-quality evidence which supports our work and demonstrated benefits of newly developed prosthetic components.

RESEARCH

Our research work focused on four main areas: outcome measurement for children and adults; development of CAD CAM system and further procedures which will allow us to make silicone partial hand prostheses as mirror copies of the non-amputated hand; problems people following upper limb amputation have at return to work; and driving abilities.

Outcome measurement

Outcome measurement has always been an important part of our clinical and research work. In children, significant correlation between UNB spontaneity and skill score as well as between the parental CAPP score and UNB test was found (16). For adults we revised the Orthotics-Prosthetics User Survey Upper Extremity Functional Scale (changed the original scoring and deleted 4 items) and ABILHAND questionnaire (changed the original scoring, selected 22 items appropriate for unilateral upper limb amputees). Both new scales are promising instruments to measure the degree of manual functioning after unilateral upper limb amputation (17, 18). In both children and adults haptic interface was tested and found to be promising for assessing upper limb function in upper limb amputees.

CAD CAM

Major Appearance and cosmesis are very important for people in many countries (5, 19). Enhanced cosmesis may imply better psychological well-being independently of body-image. Nowadays, prosthetists produce silicone partial hand prosthesis using technology where previously an individually constructed mould defined the shape of the prostheses, or with direct modelling of silicone on a model of the stump

(20, 21). With both methods the shape of the prostheses differs from the shape of the non-amputated hand. For that reason we have tried to develop a system which would enable making a prosthesis as a mirror copy of the other hand. With collaboration of two other institutions in Slovenia we have succeeded in our endeavour (22 – 24).

Return to work

Full-time employment leads to beneficial health effects and being healthy leads to increased chances of full-time employment (25). Employment of disabled people enhances their self-esteem and reduces social isolation (26). Employment rates of people after upper limb amputation are lower than employment rates for general community and may even decrease with time passing from the amputation (27). Whether a person after upper limb amputation will be still able to do the same work as before the amputation mainly depends on the type of work and the amputation level (28). We found out that people who were younger at the time of amputation and had less severe phantom pain had fewer problems, and those injured at work had more problems returning to work (29). Less than half of the patients who had had a partial hand amputation were able to do the same work as before amputation (6). The subjects who had manual work and amputated more than two fingers had more problems. Less than one-third wore their silicone prosthesis at work (6).

Driving abilities

An ability to drive is important for participation. Already in some previous studies the authors have reported that people following upper limb amputation have problems with driving and need adaptations of the car (30) and approximately 25 percent found prosthesis beneficial for driving (31). We review medical records of all the people following upper limb amputation performed in the last five years and found out that most people had problems driving. They needed from zero up to four different car adaptations, 2 on average. The most frequently suggested adaptation was automatic transmission, followed by moving of the commands from one side of the wheel to the side held by the non-amputated limb. Six needed a ball on the wheel, 4 reinforced assisted steering and one was allowed to drive only with the prosthesis. There were no differences in the number and type of needed adaptations in relation to the side of upper limb amputation and the amputation level. It was not possible to compare differences between subjects using different type of prosthesis since all except two had body-powered ones. It is important that clinicians working with persons following upper limb amputation are aware of that and refer them to driving assessment.

SECONDARY PROBLEMS

There are not many articles about secondary impairments people following upper limb amputation have as a consequence of amputation. In our preliminary study of 22 subjects we found out that they had from zero (five subjects) up to four different problems (one subject), most frequently with the shoulder on the non-amputated side and carpal tunnel syndrome, both presented in half of the included subjects. Persons who used their prosthesis less had more problems and, surprisingly, the same was found in those who had been amputated more recently.

CONCLUSION

Rehabilitation of people following upper limb amputation has to be performed by a multi and interdisciplinary rehabilitation team whose members regularly asses their work and try to improve it. The team includes also the patient and a PRM specialist. There are still many areas that are not really supported by evidence but are based on experts' experience. Good new multicentric clinical studies are needed to get better evidence for our work.

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Unless there are six authors or more give all authors' names; do not use "*et al.*". Capitalize only the first word in a paper title, except for proper nouns and element symbols.

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ASSESSMENT OF CAPACITY FOR MYOELECTRIC CONTROL – EVALUATION OF TASK DIFFICULTY, NEWLY MERGED ITEMS AND REDEFINED RATING SCALE

Helen Lindner, OT, MSc, [1,2,]
Liselotte M. Norling Hermansson, OT, PhD [2]

[1]*From the School of Health and Medical Sciences, Örebro University, Sweden,*
[2] *Centre for Rehabilitation Research, Örebro University Hospital, Sweden*

INTRODUCTION

Task difficulty and validity

The *Assessment of Capacity for Myoelectric Control* (ACMC) is an observational assessment designed to assess a prosthesis user's ability to operate a myoelectric prosthetic hand in a bimanual task [1]. The prosthesis user is encouraged to select a bimanual task for the assessment. Concern has been raised over the tasks being used in the assessments [2]. Would a prosthesis user receive different ACMC scores on different tasks?

Bimanual tasks are mostly used in ACMC assessments. An ACMC rater observes the prosthesis user operates the hand during the task performance. The ACMC total score shows the user's ability to operate a prosthetic hand as defined by the ACMC items. A detailed description of the ACMC and its psychometric evaluations can be found elsewhere [1-3].

Although the reliability and validity of ACMC have been established in upper limb prosthetic users, the difficulty of tasks being used in ACMC assessments have not been examined yet. Task difficulty in ACMC is defined as the difficulty to handle different task objects and different task steps with a myoelectric prosthetic hand. Would it be more difficult to use a myoelectric hand in a task with heavier objects and hence get a lower ACMC score? Would a user receive a lower ACMC score in a task that contain more task steps? If different tasks can lead to different ACMC scores, then a change in ACMC scores can be due to task differences, not due to an improvement in prosthetic skills.

Thus, the main objective of this study was to estimate the difficulty level of several bimanual tasks that are used in ACMC assessments. The chosen tasks will have different types of objects and the steps to perform the tasks are also different. Rasch analysis [4], a mathematical technique that estimates the difficulty of tasks based on the difficulty of ACMC items in each task, will be used. Rasch fit statistics will be use to evaluate the validity of these tasks, i.e. if the tasks are appropriate to be used in ACMC. Furthermore, it is important to investigate whether all the tasks can be used for gender, all ages and both prosthetic sides. Thus, differential item functioning (DIF) will be performed to examine if any ACMC item consistently function differently in a particular task for a particular age group, gender and prosthetic side.

Merging items and re-defined rating scale

In the evaluation of the ACMC construct , merging of related ACMC items was suggested [3]. Thus, the number of ACMC items is reduced from 30 to 22 items. Reducing the length of a test can reduce administration time, which is good for a busy clinic environment. We have also changed the definition of category-2 based on the result from the previous analysis [3]. A Rasch analysis of 22-item ACMC the re-defined rating scale would thus provide us information about the functioning of merged items and the re-defined rating category.

Thus, the first aim was to examine (i) the difficulty of tasks in ACMC assessments, (ii) the validity of the tasks, (iii) if the item functioning in each task is influenced by gender, age, and prosthetic side. The second aim was to assess (iv) the functioning of the 22-item ACMC and newly merged items, (v) the use of the re-defined rating scale.

METHOD

Subjects

A sample of 58 upper limb prosthesis users was recruited from the Limb Deficiency and Arm Prosthetic Centre (LDAPC), Örebro University Hospital, Sweden. Subjects' demographics are shown in Table 1.

Development of tasks

The development of tasks was carried out in four stages. In January 2009, ACMC raters from different countries (n=52, male=5) were asked to suggest three tasks that they would normally use in their training or assessment. The task suggestions are summarized in Table 2.

Table 1: Demographic characteristics of subjects

Subject Characteristics	N=58
Mean Age (range)	20 (2-72)
Median	13
Gender	
Male	31
Female	27
Age group	
≥ 6 years old	10
7 to 15 years old	24
≥16 years old	24
Cause of absence	
Congenital	48
Trauma	9
Illness	1
Prosthetic side	
Right (unilateral)	20
Left (unilateral)	36
Both (bilateral)	2
Prosthetic level	
Unilateral	
Shoulder disarticulation or above elbow	3
Below elbow	49
Wrist disarticulation	4
Bilateral	
Below elbow on both sides	1
Above elbow (left) & below elbow (right)	1
Prosthetic experience	
Unilateral	
< 1 year	7
1 to 4 years	14
≥ 5 years	35
Bilateral	
< 1 year on both sides	1
< 1 year (left) & >5 years (right)	1

The second stage was to select tasks for this study. The selection was based on six criteria: i) tasks can be performed with a prosthetic hand, ii) tasks of functional relevance that challenge the use of a prosthetic hand, iii) tasks that can be adapted to different ages; (iv) tasks can be performed at a clinic, v) tasks can be accomplished within 10 minutes, (vi) all ACMC items can be observed in the performance of the tasks. Six tasks were selected: *packing suitcase, mixing a ready-mix food product, sorting mail, repotting plant, setting table* and *assemble a ready-made product.*

The third stage was to write the detailed task steps and find materials for each task. Task versions for different age groups were created. The fourth stage was to pilot-test the tasks. This was to see the time needed to perform the tasks and, if all the ACMC items can be observed.

Table 2: Task suggestions

Self-caring	Household/Transportation	Construction/hobby
• Dressing • Brushing teeth • Eating and drinking	• Changing car tires/car oil • Installing smoke alarm • Grocery shopping/using wallet • Stocking groceries in shelves • Making simple food or drinks • Changing bed • Dishwashing • Sorting mail • Ironing • Setting table • Hanging laundry • Washing small laundry items • Packing suitcase • Setting up curtains • Driving	• Making clipboard, birdhouse, coat rack, or putting together a small furniture • Painting • Hanging up pictures • Sewing • Fishing • Repotting plants • Pitching a tent • Wrapping gift • Making handcraft • Playing doll dress-up • Knitting

Instrumentation

The ACMC consists of 22 items assessing six quality aspects in prosthetic control: *the need for arm support, choose the right grip strength, show good timing, use in different positions, repetitive grasp and release of objects, the need for visual support and coordination between the hands*. During an ACMC assessment, an ACMC rater takes notes of all the observable prosthetic actions performed by the prosthesis user during the task performance. The ACMC rater then scores the 22 items, using a 4-point rating scale.

Procedure

Each subject was asked to perform three tasks during one visit. An allocation technique 'Minimization' was used to assign three tasks to each subject [5]. This was to minimize the differences between the subjects' characteristics in each task. The characteristics we would want to be similar in each task were: gender, prosthetic side, prosthetic level, prosthetic experience and age.

All subjects performed the tasks in the kitchen or in the training room at the centre. Each subject had around 10 minutes break between each task. The subject or the occupational therapist decided randomly which task to perform first. The task performances were videotaped.

Since an ACMC assessment focuses on one prosthetic hand at a time, 56 unilateral users gave 168 assessments and 2 bilateral users gave 12 ACMC assessments. The 1st author watched and scored all the task videos with the ACMC. The scoring started with one task from one subject at a time. Then the 1st author selected another video from another subject, usually the same task. This was to avoid scoring the same

subject in three tasks and hence gave similar scores to the three tasks. All the scores were written down on the ACMC scoring sheets. The 1st author consulted the 3rd author for advice if there was any doubt about the scoring.

Data analysis

Rasch analysis was carried out using WINSTEPS 3.72. Item difficulty measures and person ability measures were constructed using the ACMC raw scores. Task difficulty measure is the average difficulty of the task-items for the task.

Task validity was examined by the mean-square (MnSq) statistics. Infit and outfit MnSq and were used to examine any measurement disturbance occurred in each task. The range for an acceptable goodness-of-fit is between 0.6 to 1.3 MnSq was selected. A task that shows an acceptable goodness-of-fit is considered as an appropriate task for an ACMC assessment. Differential item functioning (DIF) was performed to examine any item consistently function differently in a particular task for a particular age group, gender and prosthetic side.

The person-item map was used to assess the alignment between the subjects and the 22 items. The distribution of the items shows the difficulty of the newly merged items relative to the existing items.

The rating scale was examined by (i) the "Frequency of Use" of each category, (ii) "Person Measures" for each category, which should increase from a category representing low ability to one representing high ability, (iii) "Threshold Measure" between any two ratings. This should also increase with increasing category number.

RESULT

Task difficulty and validity

Based on this sample, *packing suitcase* (-0.26 logits) is the easiest task (Table 3). *Assemble a ready-made product* and *setting table* are equally difficult (0.13 logits). The difficulty range is -0.26 to 0.13 logits, i.e. 0.39 logits difference. From the ACMC raw score to logits conversion table in Winsteps, a change in 0.5 logits is equivalent to 2 ACMC raw scores. Hence, a change in 0.39 logits is less than 2 ACMC raw scores, suggesting that the impact of the task difficulty difference on the ACMC score is minimal.

The Infit and Outfit MnSqs are all within the acceptable range (Table 3), implying that these tasks are appropriate to be used in ACMC assessments.

No item exhibit DIF in gender and prosthetic side, implying that the tasks are appropriate for both genders and both prosthetic sides. Two items 'holding without visual feedback' and 'holding in motion without visual feedback' exhibit age DIF. These two items are relatively easier for those who are age 6 or younger.

Table 3: Task difficulty measures and task fit statistics

Task	Mean age (yr)	Difficulty (in logits)	Infit MnSq	Outfit MnSq
Packing a suitcase	18.1	-0.26	1.15	0.88
Sorting mail	20.0	-0.13	0.81	0.63
Mixing a ready-to-eat product	18.5	0.05	0.94	0.73
Repotting plant	19.7	0.09	1.15	0.88
Assemble a ready-made project	19.7	0.13	0.77	0.66
Setting table	22.1	0.13	0.92	0.70

The 22item ACMC with newly merged items

The person-item map in Fig.1 clearly shows the distribution of subjects in relation to the ACMC items. The mean person ability is 2.34 logits (mean item difficulty is set at 0 by default), indicating that this sample has a high ability in operating a myoelectric prosthetic hand. Since 35 out of 58 subjects have more than 5 years of prosthetic experience, a high mean ability is expected.

The newly-merged items are circled (Fig.1). Their positions along the vertical scale are similar to their positions in the 30 item ACMC [2], indicating they are functioning as expected.

The redefined rating scale

The use of 4 rating categories is shown in table 4. The observed person measures increase as the category increases. The use of category-2 is slightly lower than we expected (22%). In Fig.2, however, it shows that the probability of selecting category-2 is the same as selecting category 1, indicating that the new category-2 definition has improved the functioning of the rating scale.

DISCUSSION

Using the ACMC items, which measure how a prosthesis user grasps, hold and release different objects, the six tasks have similar difficulties. This may be surprising for the readers since; in general, as in our knowledge, some tasks are more difficult than other tasks. However, the ACMC is not designed to measure how well a prosthesis user performs a task. The ACMC items do not measure, for example, if the plant is not straight after repotting, or if the milk is spilt on the table during mixing the food product.

The ACMC items measure, for example, if the user is able to grasp the plant with the prosthetic hand using the right grip force, or, if the user is able to maintain holding a shoe bag when putting shoes inside. In the task 'packing a suitcase', a few prosthesis users dropped the shoe bag when putting shoes inside. This was because the users did not increase the grip force when the shoe bag got heavier after putting the shoes in it.

It is interesting to find that children got higher scores in the two items that measure holding without looking at the hand. When children are engaged in these tasks, they only focused on mixing the dough or putting soil into the pot, not on their hands. Adults, who were fully aware of the video cameras, looked at their prosthetic hands more often than children and hence received lower scores on these two holding items.

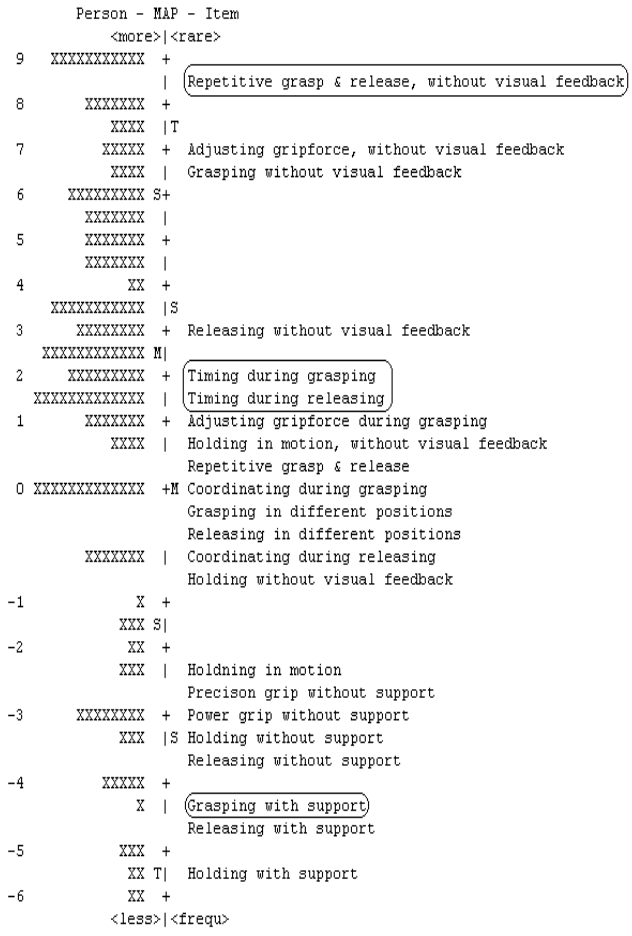

Figure 1: Relationship between ACMC items and the subjects

X= subjects, M=mean, S=1 standard deviation (SD) from the mean, T=2 SD from the mean

Table 4: Summary statistics for the 4 rating categories

Category	Frequency of use (%)	Person measure	Threshold measure
0- not capable	616 (16)	-5.38	None
1-somewhat capable	642 (16)	-1.20	-3.02
2-generally capable	882 (22)	1.69	-0.01
3-extremely capable	1776 (45)	6.39	3.03

Frequency of use: the no. of persons being scored
Person measure: mean person ability measure in the category
Threshold measure: the difficulty measure between 2 adjacent categories

The six tasks were selected from ACMC raters' suggestions. Many suggestions are suitable for ACMC assessments. However, because of the selection criteria, for example, can perform at a clinic, suitable for all ages and takes only 10 minutes, limited the task choice for this study. These six tasks are now standardized and are suitable for retest purpose. A change in ACMC scores between different test occasions from these tasks may indicate an improvement in prosthetic skills or even a change of device.

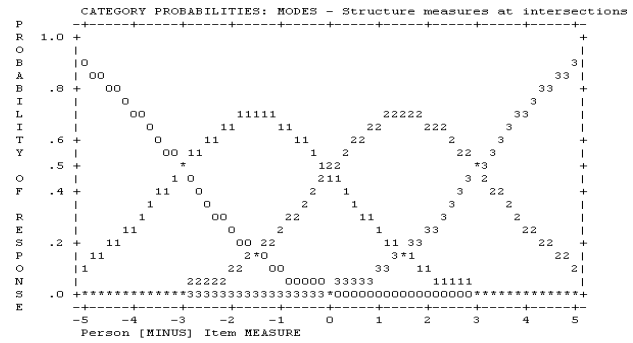

Figure 2: The probability curves of the 4 rating categories. The 0, 1, 2 and 3 curves represent the 4 categories

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A QUANTITATIVE OPERATIONAL PERFORMANCE MEASURING SYSTEM FOR A MYOELECTRIC HAND: A PRELIMINARY STUDY

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INTRODUCTION

We are developing a quantitative measuring system for basic operation abilities of myoelectric prosthetic hands. Preliminary results for the prosthetic hand users are shown in this report.

While there are numerous activities for developing dexterous---multi functional---prosthetic hands, most commercial electric powered prosthetic hands are limited to single active function systems. Such commercial prosthetic hand users, however, have abilities to perform various daily or work related activities effectively, when their prostheses are appropriately fitted.

Control methods of the commercial prosthetic hands were categorized into three distinct generations by a previous work [1]; first generation used an on-off switch type control scheme for hand motor activation. Second generation hands have ability to adjust thresholds for motor activation, and proportional controllability of motor speed was provided in this generation. In the third generation, control options can be modified easily, because they utilize programmable microprocessors. Even in the third generation or proportional control systems, motor activation is based on the threshold of controller input signal; therefore, appropriate adjustment of the threshold, or an amplifier gain in a myoelectric sensor, is a significant issue for high-performance uses of the prosthetic hand, which may leads to high acceptance ratio of myoelectric hand prostheses.

Performance of prosthetic hand use is also affected by clinical training: myoelectric signal training, control training and functional training. [2] Our target is the control training stage, in which both of ability to control remnant muscles and socket-sensor fitting have effects on myoelectric control; and therefore, it is crucial for clinicians to evaluate performance in this stage.

Such performance can be evaluated in terms of basic and functional operation ability; the functional ability is measured by various methods, such as a required period or quality of task completion with the hand prosthesis. [3][4] The basic operation ability relates to how the user can control the hand open-close function as they intend to, and this is affected by the adjustment of the threshold or the amplifier gain.

METHODS

Figure 1 shows a configuration diagram of the measuring system, which consists of a personal computer (PC), a MyoBoy® (Otto bock HealthCare) and two myoelectric sensors. Two software tools---a "Switching Evaluation Tool" and a "COM Wrapper"---are installed on the PC. The first tool was developed to measure basic operation performance of a human-machine interface device (HID). The second tool receives data sets from the MyoBoy® and produces HID events--- keyboard inputs or mouse button clicks---that are used by the measuring tool.

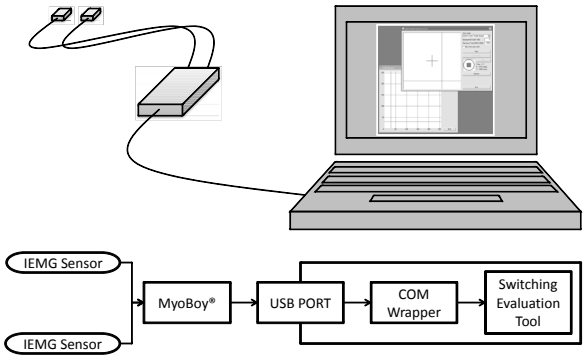

Figure 1: A measuring system configuration diagram.

COM Wrapper

Figure 2 shows a window image of the COM Wrapper operation. The COM Wrapper reads a COM-port of the PC to obtain output data patterns from the MyoBoy®. Received data patterns are plotted on a two-dimensional graph as shown in Figure 2. There are two levels of thresholds for each axis. These thresholds correspond to the thresholds for the myoelectric prosthetic hand activation; when the myoelectric sensor output voltage exceeds one of the lower thresholds, the

hand motor begins to rotate. (Note that the higher thresholds are used for 'four-channel' mode of MyoBock® system, and the higher thresholds are not used in this paper.) In a same way, the COM wrapper produces HID events that are used by the measurement tool. The HID events for these thresholds are defined by a configuration file.

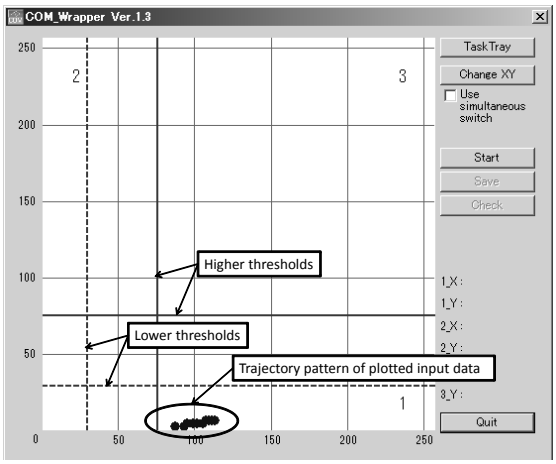

Figure 2: An example of a window image of the COM Wrapper operation.

Switching Evaluation Tool

Developments of the "Switching Evaluation Tool" started in 2006, in order to conduct engineering evaluations of a general purpose myoelectric switch interface. [5] The tool is an application program running on a Windows™ (Microsoft Corporation) operating system.

An external switch interface device, which translates switch operation into HID events, is required to measure switch operation ability, but in this paper, the COM Wrapper deals with this function.

This tool evaluates operation ability of switch-type interface in terms of 'Quickness', 'Timing controllability' and 'Sustainability', as listed below.

1. Quickness

   ① Switch Close Delay: Response time for a switch closure.

   ② Switch Open Delay: Response time for a switch opening.

   ③ Switch Repetition Time: Required period for switch close-open repetitions. The number of repetition time is defined by a configuration file. The default number is ten.

   ④ Switch Repetition Time (2ch): The 'Switch Repetition Time' with two switches alternation.

2. Timing Controllability

   ① Switch Close Timing Spread: Variation in switch close timings.

   ② Switch Open Timing Spread: Variation in switch open timings.

   ③ Switch Repetition Timing: Variation in close-open repetition timings. The number of repetition time is defined by a configuration file. The default number is ten.

   ④ Switch Repetition Timing (2ch): The 'Switch Repetition Timing' with two switches alternation.

3. Sustainability

   ① Switch Endurance Period: Time period for sustaining switch closure.

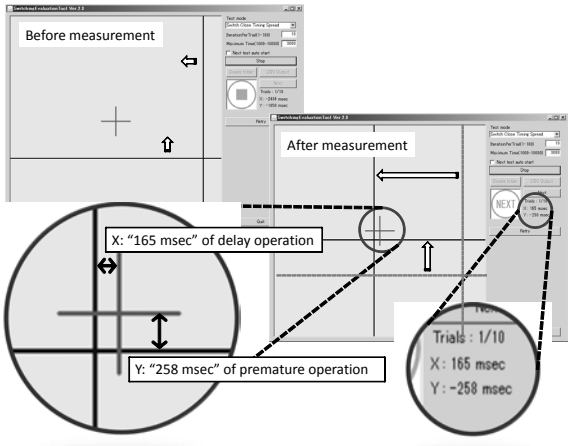

Figure 3: An example of the measurement of the timing controllability with the switching evaluation tool.

Figure 3 shows an example of a measurement with the switching evaluation tool. The tool window consists of two areas, a measuring area (left) and an operation area (right). In the measuring area, there are a target cross-shape at the centre, a vertical line and a horizontal line. The operator is asked to align both vertical and horizontal lines to overlap the centre of the cross-shape.

The operation procedure is similar to that of a claw vending machine or a toy crane machine. The example in Figure 3 shows a measurement of the "Switch Close Timing Spread", where we can estimate switch operation timing controllability from distances between the target cross-shape and the vertical or horizontal line.

These lines initially locate at arbitrary distance from the centre, as shown in a left panel of Figure 3. When output voltage of the myoelectric sensor exceeds the threshold, the COM Wrapper sends the corresponding HID command to the measuring tool, and one of the lines begins to move toward the centre. The operator can relax after the line movement started. Then the operator needs to predict the timing when the line reaches to the centre, and to contract the muscle again, in order to stop the line movement. In the case of measuring the "Switch Open Timing Spread", the operator requires to continue muscle contraction during the line movement, and to relax the muscle for stopping the line motion.

Measurements for the "Switch Close Delay" or "Switch Open Delay" proceed in a similar way, but the line motion are hidden before the line reaches the centre, and therefore, the operator cannot predict line motion, as shown in Figure 4. When the line reaches the centre and the operator finds the line movement, the operator needs to stop line movement. Reaction time can be calculated from the distance of lines and the target cross-shape.

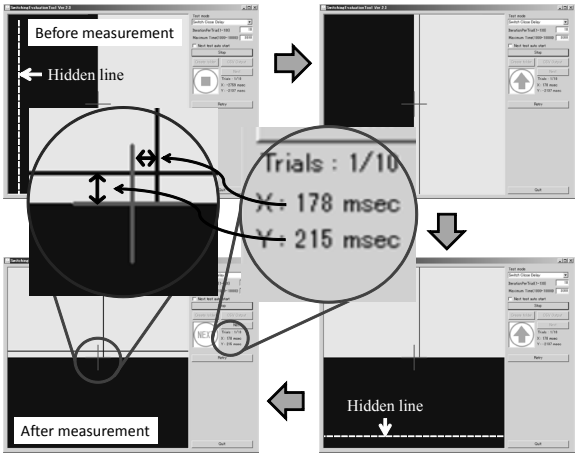

Figure 4: An example of the measurement of the response times with the switching evaluation tool.

By using this tool, we have conducted measurements to a muscular dystrophy patient group and sound volunteer participants through 2008-2009. In these measurements, we used mechanical switches and the myoelectric switch; results indicate that the response time of the muscular dystrophy group tends to increase, and the myoelectric switch shows shorter response time than mechanical switches. [6]

RESULTS

Participants for this preliminary study are a user of a forearm prosthetic hand (MyoBock® digital hand) and a user of an upper arm prosthetic arm (hybrid system of a MyoBock® DMC hand and a cable control elbow). Both of the participants have over five-year prosthetic use experience, and written informed consents and agreement of a local ethics review board was obtained prior to the measurements. We measured "Switch Close Delay" and "Switch Close Timing Spread" for both participants, and "Switch Open Delay" and "Switch Open Timing Spread" were obtained only from the second participant.

Figure 5 and 6 shows measured data pattern distribution for switch closures and switch openings, where X-axis is results for response time and Y-axis for timing controllability. Each plot is the average of twenty-time repetition. In these graphs, "HAND_A" and "HAND_B" represent the first and second participants; "inner" and "outer" are sensor locations on the remnant part of the limb. "Non-disabled volunteer Group" and "Muscular dystrophy Group" are results with mechanical switches from previous research projects. These results show that the participants for this preliminary study show high response speed and high timing controllability, and there are no considerable differences in response speeds between myoelectric hand users and the non-disabled volunteer group.

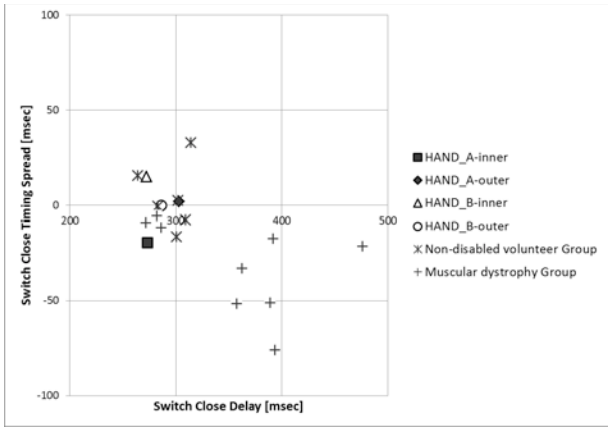

Figure 5: Measured data distribution of switch closures.

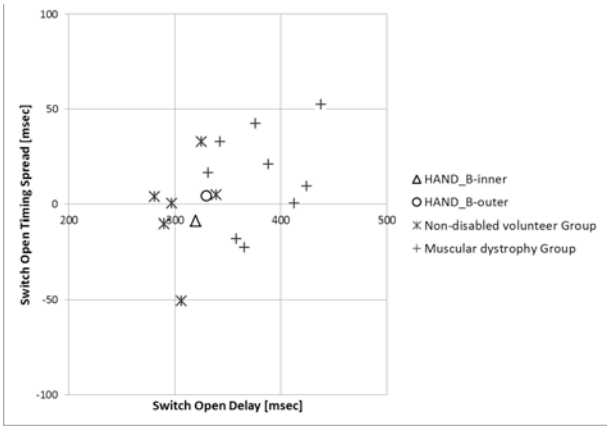

Figure 6: Measured data distribution of switch openings.

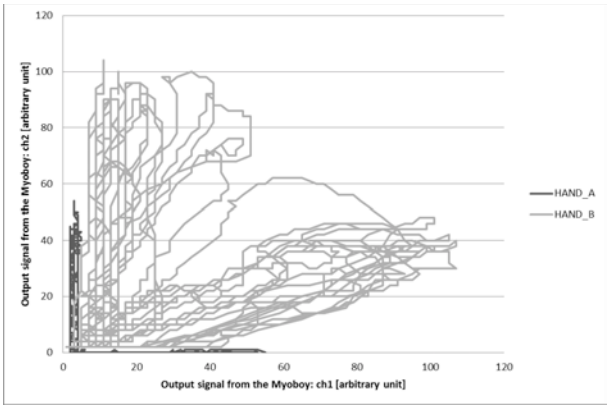

Figure 7: Distribution diagram of output data patterns of the MyoBoy®.

The reason why we did not measure "Switch Open Delay" and "Switch Open Timing Spread" for the first participant is that he is the user of the digital hand system. Figure 7 shows an example of a distribution diagram for output data patterns from the MyoBoy® for both of the participants. It is clear that the distribution range for the HAND_A (digital hand user) is smaller than that for the HAND_B (DMC hand user). In the case of the digital system; when the myoelectric sensor output exceeds the lower threshold, the hand motor is activated with a constant speed. Therefore, the user does not need high myoelectric sensor output and this makes difficult for the digital hand user to keep high signal intensity, which are required for measuring the "Switch Open Delay" and "Switch Open Timing Spread".

DISCUSSION AND FUTURE WORKS

This report introduced a measuring tool for basic operation abilities of myoelectric prosthetic hands. Results indicate that the participants for these preliminary measurements have high response speed and high timing controllability, and we believe that these factors have much impact on satisfaction or acceptance ratio.

From the results in Figure 5 and 6, some participants in the muscular dystrophy group showed long response times (400 or 500 milliseconds); but even with such long response times, they showed better timing controllability than we expected. This means that they compensate the response delay by using higher brain function of prediction. It is possible that this is not the case for the muscular dystrophy patients group but the prosthetic hand user group. When prosthetic user candidates have long response time; even if they can operate the prostheses well, they may feel mental burden to use their prostheses.

This preliminary study is limited to the measurements of two myoelectric hand users, and both of the users have long experience of the prosthetic use. It is required to make further measurements for prosthetic hand users and other patients. These tools are distributed at some hospitals in Japan from this year, and we believe that the tools support their daily activities.

ACKNOWLEDGEMENTS

A part of this work was supported by the Special Coordination Funds for Promoting Science and Technology of Japan Science and Technology Agency, and the Health Labour Sciences Research Grant of the Ministry of Health, Labour and Welfare of Japan.

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CHALLENGES AND SOLUTIONS IN CONTROL SYSTEMS FOR ELECTRICALLY POWERED ARTICULATING DIGITS

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INTRODUCTION

The invention and clinical application of electrically powered and independently articulating digits is relatively new in the field of external upper limb prosthetics. When utilized for patients with amputations or absence at the partial hand level, these components offer the potential for a range of active functional grasping patterns that were unavailable with previous technology. Their application and examples of their use have been documented by various authors.[1,2]

The introduction of these systems is accompanied by the challenge of controlling them. Any electrically powered prosthetic system requires a method of concise, deliberate, and repeatable control be implemented in conjunction with focused therapy in order to be successful. Traditional control schemes of prosthetic devices for more proximal levels of absence are less straightforward when applied to an electrically powered partial hand device. Space constraints, limits of myoelectric input, the desire to maintain available residual anatomy range of motion, and complexity of potential prosthetic motion make the control of these systems particularly challenging. Integrating novel and creative systems of control with therapy will enhance the function of these systems for each user.

RESIDUAL HAND PRESENTATIONS

"The primary goals of amputation surgery are preservation of length and useful sensibility, prevention of symptomatic neuromas and adjacent joint contracture, early prosthetic fitting where applicable, and prompt return of the patient to work or play"[3] The surgical principals guiding those performing amputations within the hand dictate the levels of amputation and generate a wide variety of partial hand presentations. This variety poses different challenges based on remaining anatomy and available range of motion and input sites.

Manufacturer's component systems allow for fitting of digits at different amputation levels based on prosthetic digit build height and overall length. There are certain constant indications for application of articulating digits however. One indication is the complete loss of at least one and up to five digits. No manufacturer currently offers an electric partial finger prosthesis that can be fit when a substantial portion of the finger remains.

Another indication is a majority of the carpal bones remaining. The remaining carpus allows for the potential of active wrist motion. Wrist motion is well understood to be critical for functional activities and the lack of motion typically results in compensatory movements strategies. The advantages of even passive prosthetic wrists have been documented.[4] Preservation of active wrist range of motion is therefore one of the essential design criterion of a partial hand prosthesis.

The thumb is the most important digit in its contribution to grasp and pinch. In the prosthetic treatment of any partial hand presentation where the thumb remains it is absolutely essential for the prosthetic system to allow as much active range of motion as is available. Residual finger anatomy should also be evaluated and in any case where useful motion is available should remain as unhindered as possible. In addition, therapy to improve thumb and finger strength as well as range of motion is critical to improve outcomes.

The restriction of physiological range of motion within a prosthetic system can come from limitations imposed by the structure of the prosthesis. Socket or frame trimlines, pressures within the socket, or even the materials used can affect the user's ability to move. Restriction to motion can also be imposed based on positioning of the inputs to the control system. This occurs when undesirable or unintended prosthetic operation is elicited by motions that could otherwise be beneficial. For example, the location of electrodes for myoelectric control in transradial and wrist disarticulation systems is most commonly on the remaining wrist and finger flexors and extensors. When this same control scheme is applied to a partial hand system where wrist motion remains, the control of the fingers is directly tied to wrist flexion and extension. This presents a functional deficit and most certainly an obstacle to a successful outcome even with intensive therapy.

MIRCROPROCCESOR CONTROL SCHEMES

Commonly electric prosthetic systems are designed with myoelectric control of the various components. The preferred myoelectric control scheme consists of dual site agonist and antagonist muscle pairs. Each site controls one degree of freedom for the component being controlled: hand, wrist rotator, elbow, etc. A switching mechanism, either myoelectric or electro-mechanical, is used to change the component currently operating. Generally these schemes are well understood by rehabilitation professionals and training for prosthesis control is straightforward.

Another more challenging scheme is single site control. A single myoelectric input is utilized to operate all functions of the prosthesis. Microprocessors determine the direction of motor motion based on differing algorithms.[5] Alternating motion, rate dependant direction, and automatic closing are all examples of single site control methods. When more than one component is controlled another switch is necessary to move between components.

Rarely is prosthesis function as fluid when using single site control as compared to dual site. The necessity for additional switching or the inability to change directions seamlessly creates delays in control. However, when other input options are not logical or unavailable, the ability to utilize one input for control can be vital to the success of the system.

In addition to electrodes used for myoelectric control, alternate inputs can be used. Force sensing resistors (FSR's), linear transducers, and strain gauges can all be used for proportional input into control systems.[5] These devices can be used in conjunction with or as replacements to myoelectric inputs. In some cases three input systems are advantageous where the third input, typically an alternate input, has direct control over a component or is used as a switch.

Pattern recognition for myoelectric control is an emerging technology that is very promising. With the application of pattern recognition, isolation of individual muscles becomes much less important or problematic. The requirement of signal separation in dual site control is diminished and the potential for control of more degrees of freedom is gained.[6] Undoubtedly when it becomes available to the prosthetics industry there will be applications for the powered partial hand prosthesis.

The considerations in prosthesis design provide a basis for socket design but also for input determination and prosthesis control. Each partial hand presentation has its own limitations and challenges but also provides different opportunities for control.

CONTROL SYSTEM CONSIDERATIONS

5 fingers absent

The prosthesis for the residual limb with all fingers missing is arguably the easiest to control of any those discussed. The lack of fingers allows for the preferred method of dual site myoelectric control with intrinsic muscles of the hand. The most logical when available are the hypothenar and thenar eminences due to their size and typical signal separation.[7]

The simplicity of control is balanced by the decrease in residual hand function due to lack of a thumb. The lack of any digital sensation severely complicates training and function. One decision that must be made is whether or not to motorize the prosthetic thumb but the control of either system is likely to be similar.

4 fingers missing – thumb remaining

The presence of a thumb makes the use of the thenar eminence as a myoelectric site inadvisable as thumb motion and prosthesis function would be linked. This would create a conundrum of control for the user. Hypothenar musculature is still very viable for control with this presentation.

For dual site intrinsic myoelectric control the second site is likely the lumbricals or dorsal interossei. Imagined 2nd-5th digit MP flexion along with PIP and DIP extension generally results in the best signal for these groups. This can be described as having the patient "fold" the hand at the knuckles while keeping the fingers straight. These small intrinsic muscles are viable contributors of myoelectric signal but certainly require training to have sufficient stamina and strength to be used functionally.

Dual site control is still the preferred method of control, but only if achievable, consistent, and functional. Single site control is a viable option at this level due to the relatively strong and isolated hypothenar muscles.

3 fingers missing – thumb and 2nd digit remaining

Having the 2nd digit and thumb with active range of motion and sensibility provides what the previous levels do not have: true pinch and grasp native to the residual hand. Fine dexterity is typically not an issue. Powerful and stable grasp of larger items, however, can still be a challenge as grasp is limited to a "ring" created by the two digits.

Myoelectric input is certainly available from hypothenar muscular if present. When attempting to find a second myoelectric site again the small intrinsic muscles of the lumbricals and dorsal interossei are candidates. Intensive training to separate 2nd digit motion from this second site is imperative in order to have this be a successful outcome. Due

to the complexity involved in training isolation of 2nd digit motion from activation of the dorsal interossei and lumbricals the potential for control error is great. Single site control does not have these same issues and is less error prone but lacks the responsiveness of dual site control.

3 fingers missing – thumb and 5th digit remaining

This presentation is the most difficult to address from a control standpoint as the only intrinsic myoelectric site available that doesn't involve the thumb or 5th digit are the 1st and 2nd dorsal interossei and lumbricals. Only single site myoelectric control is achievable with intrinsic musculature and this is by no means straightforward due to the motion requirements of the two remaining digits. Dual site myoelectric control can be achieved through use of one electrode on the 1st interosseous with a second electrode on wrist extensor compartment. This does indeed tie prosthetic finger opening with wrist extension but separating the finger closing signal out minimizes the functional deficit. Therapy to train in the use of these motions for function is critical to a successful outcome.

NOVEL SOLUTIONS

The four partial hand presentations and potential control schemes discussed offer real solutions. However, being creative with control schemes and inputs, in conjunction with directed therapy, can reward the user with improved control.

One such potential solution is to the quandary of intrinsic dual site myoelectric control. In cases except that of 5 fingers missing, dual site control is achieved with at least one site being of smaller interrelated muscle groups. Retaining one myosite over the hypothenar eminence and replacing the second input with an alternate input can significantly reduce the crosstalk associated.

A technique utilizing a FSR has been used with good success. (fig 1) In a prosthesis design with flexible socket and rigid frame, the socket has been extended proximally on the dorsal aspect of the wrist. This socket flexes with the patient during wrist extension. The rigid frame is adjusted to terminate distal to the wrist crease thus not interfering with wrist motion. Placing a FSR between the socket and frame creates pressure on the sensor during a defined amount of wrist extension.

The benefit of such a solution is to allow for very repeatable and reliable dual site control of the prosthesis without placing electrodes on more proximal muscles. The FSR produces a signal during wrist extension but the degree of extension at which the FSR is triggered is adjustable. Thus opening of the fingers can be reserved for the last 10 or 15 degrees of motion. This minimizes the functional deficits of prosthesis function related to residual joint ROM.

Similarly a linear transducer can be used as the alternate input. By anchoring the transducer above the wrist and to the dorsal aspect of the frame, wrist flexion can be captured to produce input signal. As with the FSR, the point at which the signal is produced is adjustable.

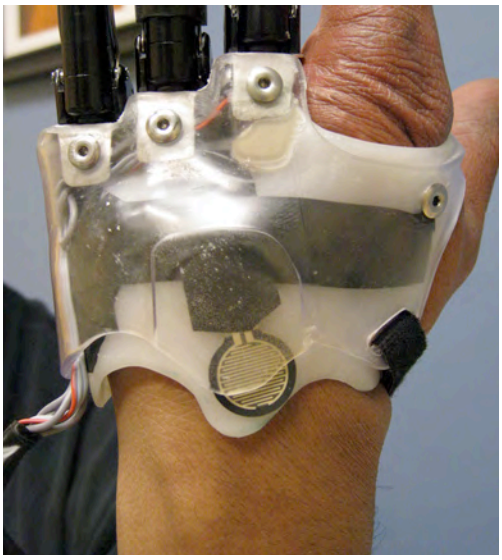

Figure 1: example of FSR placement

CONCLUSION

Electric partial hand prostheses with individually articulating digits are currently being fit. The variety of residual limb presentations creates numerous challenges of control for these complex systems. By utilizing and training the user in innovative control schemes improved control of the prosthesis can be achieved.

ACKNOWLEGEMENTS

I would like to acknowledge Robert Dodson CPO, Shawn Johnson OTR/L, and Amy Nistico OTR/L for their contributions to the content of this paper as well as their editorial assistance.

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COMPARISON OF TWO MYOELECTRIC MULTI-ARTICULATING PROSTHETIC HANDS

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INTRODUCTION

Myoelectric prosthetic hands with powered multi-articulating fingers offer users the promise of increased functional grasp options, previously unavailable in a myoelectric prosthetic hand. We take a comparison look at two multi-articulating myoelectric prosthetic hand systems, the bebionic v2 from RSLSteeper and the i-LIMB Pulse from Touch Bionics, Inc.

These myoelectric prosthetic hands are categorically the same design, given that each hand has five multi-articulating powered fingers, including a thumb that can also be passively positioned in an opposed or non-opposed manner. By taking a closer look at each system, it is apparent that there are some unique features that differentiate these prosthetic hands. This comparison will focus on discussing the prosthetic hand features including available grip patterns, functional programming, graphic user interface, component compatibility, available load testing/grip force summary, battery options, glove options and sizing. In conclusion, the participant reader will have a better overall understanding of these systems and be better prepared to make a component decision regarding the desired clinical outcome for their clients.

AVAILABLE GRIP PATTERNS

With individual motors for all five fingers, each hand has the ability to achieve four selected, pre-programmed grip patterns. Table 1 shows the available pre-programmed grip patterns for each hand.

The i-LIMB Pulse has 10 available pre-programmed grip patterns to choose from while the bebionic v2 has 11 available pre-programmed grip patterns. The 3 Jaw Chuck grip pattern provides prehension when the thumb, index and middle fingers close together. The i-LIMB Pulse flexes the 3rd and 4th fingers fully closed to remain out of the way, while the bebionic v2 is designed so the 3rd and 4th fingers move with the thumb, index and middle fingers. Power grip provides prehension with the bebionic v2 by closing the index, middle, 3rd and 4th fingers onto an object followed by the thumb closing down over the dorsum of the index and middle fingers to secure the grip. With the i-LIMB Pulse, power grasp is not pre-programmed, but can be achieved by using a manual stall technique with the thumb. Once the fingers have made contact with an object, the opposed thumb is allowed to flex close to secure the grip pattern.

Table 1: Available Hand/Grip Positions

Hand/Grip Positions	Multi-Articulating Hands	
	i-Limb Pulse	*Bebionic v2*
3 Jaw Chuck	(1)	(1)
Power Grip	*	(2)
Hook grip	*	**
Lateral/key grip	(2)	(3)
Index Point	(3)	(4)
Natural Hand	(4)	(5)***
Standard precision pinch open	(5)	n/a
Thumb precision pinch open	(6)	(6)****
Standard precision pinch closed	(7)	n/a
Thumb precision pinch closed	(8)	(7)*****
Thumb park continuous	(9)	n/a
Thumb park quick	(10)	n/a
Pinch Grip	n/a	(8)
Trigger Grip	n/a	(9)
Column Grip	n/a	(10)
Mouse Grip	n/a	(11)
Finger Adduction	n/a	******
Open Palm	*******	*******

*achieved using a manual stall of thumb during flexion

**achieved in power grip at the start of thumb flexion

***called relaxed hand position

****called precision open grip

*****called precision closed grip

******non programmed grip pattern, achieved between index and middle fingers or middle and 3rd fingers as they flex closed, best in power grip, key grip and 3 jaw chuck

*******non programmed grip pattern, achieved with hand fully opened with thumb non-opposed

The Lateral/Key grip moves the index, middle, 3rd and 4th fingers to a partially closed(bebionic v2) or fully closed(i-LIMB) position, at which time the thumb can be opened/closed against the index finger. Index point is a grip pattern where the middle, 3rd and 4th fingers are flexed closed with the non opposed thumb closed against a fully extended index finger. The i-LIMB "Natural Hand" is a grip pattern which moves and holds the fingers in a slightly flexed, anatomically neutral position. This grip pattern is achieved with the bebionic v2 by accessing the Relaxed Hand Position grip pattern. Standard Precision Pinch Open(i-LIMB) allows the index and thumb to open/close while the middle, 3rd and 4th fingers remain positioned fully extended. Thumb Precision Pinch Open(i-LIMB) or Precision Open Grip(bebionic v2) allows the index to close against a partially flexed, parked thumb, while the middle, 3rd and 4th fingers remain positioned fully extended. Standard Precision Pinch Closed(i-LIMB) allows the index and thumb to open/close while the middle, 3rd and 4th fingers remain positioned fully flexed. Thumb Precision Pinch Closed(i-LIMB) or Precision Closed Grip(bebionic v2) allows the index to close against a partially flexed, parked thumb while the middle, 3rd and 4th fingers remain positioned fully flexed. Thumb Park Continuous(i-LIMB) moves all the digits to full extension and the thumb can be flexed or extended by the input signals. Thumb Park Quick(i-LIMB) moves all the digits to full extension and the thumb can be operated by the input signal for a time period of 1.5 seconds at which time the hand operation automatically returns to normal function for all digits. Pinch Grip(bebionic v2) is described as the opposed thumb closing to meet the closing index finger, while the other fingers close until they meet resistance or until the close signal stops. Trigger Grip(bebionic v2) is when the middle, 3rd and 4th fingers close securely onto a handle object followed by the opposed thumb flexing closed to secure the grip. The index finger can then close on the trigger of the device or open to a fully extended position before the other fingers will release their grip. Column Grip(bebionic v2) is when the non-opposed thumb flexes into the palm, followed by the flexing index, middle, 3rd and 4th fingers to form a fixed column point with the PIP aspect of the index and middle fingers. Mouse Grip(bebionic v2) flexes the non-opposed thumb and 4th finger to secure the sides of a computer mouse and uses the middle and 3rd fingers to provide additional stability. The index finger closes with a close signal and opens with an open signal to complete the mouse click cycle.

ACCESSING GRIP PATTERNS

The i-Limb Pulse can be programmed to utilize 4 grip patterns using 4 different input signals from the user. These input signals are described as hold open, double impulse, triple impulse and co-contraction. Each of these inputs can be programmed or linked to one of the 10 available pre-programmed grip patterns by using the BioSim Graphic User Interface (GUI). The i-LIMB Pulse is unaware of the thumb position relative to selected grip pattern. This means that the user must coordinate the non-opposed or opposed thumb position with the selected grip pattern. The user must also select the degree of thumb rotation desired from opposed to non-opposed endpoint positions.

The bebionic v2 hand can be programmed to utilize 8 pre-programmed grip patterns total. These include 2 in primary opposed thumb position, 2 in secondary opposed thumb position, 2 in primary non-opposed thumb position and 2 in secondary non-opposed thumb position. The bebionic v2 hand accesses the grip patterns using input signals that are first dependent on one of the two definitive thumb positions, opposed or non-opposed. When the thumb is situated in the opposed position, 3 Jaw Chuck may be programmed as the default grip pattern with a switching input causing the hand to select a secondary grip pattern, power grasp, for example. When the thumb is in the non-opposed position, Key Grip may be programmed as the default grip pattern with a switching input causing the hand to select the secondary grip pattern, index point, for example. Regardless of current grip pattern, every time the thumb is shifted to the other toggled position, the default grip pattern for that thumb position, opposed or non-opposed, is automatically selected. Switching grip patterns within the current thumb position is achieved in the following manner. The hand must first be fully opened and then within 1 second following full extension, the user must provide either an (open) impulse or a co-contraction to select the alternative grip pattern for that thumb position. With the bebionic v2 hand switched ON, selecting the primary grip patterns or the secondary grip pattern options can be achieved by pressing the program switch for less than 2 seconds. The switch can be accompanied by an audible sound and a vibration if activated on bebalance. With The bebionic v2 hand switched OFF, the user can enter glove donning mode by pressing and holding the on/off membrane switch for 4 seconds until the hand automatically moves into the glove donning position. To exit glove donning mode, the user must press and hold the membrane switch for 4 seconds until the hand moves out of glove donning mode and into the default grip pattern selected.

GRAPHIC USER INTERFACE (GUI)

The i-LIMB Pulse hand utilizes the BioSim Basic or BioSim Professional software as the GUI. The hand communicates to the GUI via a USB BlueTooth enabled connector. This allows the prosthetist the ability to analyze the patients myoelectric signals, configure the myoelectric control strategy and view or change current input triggers for selected grip patterns. In BioSim Basic, thresholds are preset, while BioSim Professional allows the prosthetist the ability to change thresholds and customize power to motors for possible "new" automatic grip patterns. BioSim also gives battery status and uses serial number recognition for desired hand connection. The i-LIMB Pulse also has an available USB BlueTooth connector called BioSim Patient. This tool allows the patient the ability to view their myoelectric input signals and change their input trigger/output grip pattern setup at any time.

The bebionic v2 hand utilizes the bebalance software as the GUI. The hand communicates to the GUI via an RF module/USB dongle connection. The hand, which houses the RF module, creates a unique connection with the GUI which allows the prosthetist the ability to view and change the setup for the hand while the system is being used by the patient. With the bebalance software open and the hand switched ON, the user simply holds the ON/OFF membrane switch depressed for more than 4 seconds to enable the RF module. The USB dongle blue light is solid, then a quick release of the ON/OFF switch completes the connection of the bebionic v2 hand to bebalance GUI. The bebalance GUI allows the prosthetist the ability to chose 1 of the 5 different operating modes, view myoelectric input signals, set and/or change ON and MAXIMUM thresholds, change default grip and second grip within allowed configuration options and as a training tool for working with the patient.

COMPONENT COMPATIBILITY

The following table creates a list of components that have been approved for compatible use by both Touch Bionics, Inc or RSLSteeper. Seeing this in one table opens up opportunities for multiple design configurations, as well as possible plug and play options with a patients existing myoelectric prosthesis. Please refer to Table 2 for specifics regarding component compatibility for each multi-articulating hand system.

Table 2: Component Compatibility

Component	Multi-Articulating Hands	
	i-Limb Pulse	***Bebionic v2***
Otto Bock Inputs: 13E125, 13E200, 13E202	?,yes,?	yes, yes, yes
Otto Bock Inputs: 9X14, 9X18, 9X25, 9X37	yes,yes,yes,yes	?,?,?,?
Otto Bock Inputs: 9X50, 9X51, 9X52	yes,yes,?	yes,?,yes
LTI Inputs: DC200B=50	yes	yes
LTI Inputs: TP01, LT01, LT02	?,yes,yes	yes,?,?
Motion Control Inputs: 3010546, 3010292	yes, yes	?,?
RSL Steeper Inputs: SEA200	yes	yes
Otto Bock Elbow: 12K44=	yes	yes
Otto Bock Elbow: 12K50=	yes	yes
LTI Elbow: BE330	?	yes
Motion Control U3, U3+	yes	yes
Otto Bock Wrist Rotator and Myorotronic	yes	yes

LOAD TESTING/GRIP FORCE SUMMARY

A unique feature to the i-LIMB Pulse is the ability to add additional grip force to the object held by the hand. Table 3, in the following i-LIMB Pulse column, displays both the initial pinch force value and the "pulse" pinch force value. It is interesting to observe the differences between each hand regarding the load testing/grip force summary as listed in Table 3. Some of the parameters are very close to each other while others, like the overall load limits for each hand differ greatly. Some of these measurements may speak to the durability of each hand system as well as when it may or may not be clinically indicated to fit one of hand over the other. Each manufacturer makes a cautionary statement that their hand is designed and recommended for mild to moderate activities. They are not recommended for heavy duty usage or for exposure to wet environments. The glove options provide adequate protection for most normal situations, however extra precautions should be made not to expose the fingers/motors to water or a wet environment.

Table 3: Load Testing/Grip Force Summary

Specified Load/Grip Force Parameters	Multi-Articulating Hands	
	i-Limb Pulse	*bebionic v2*
Lateral pinch force	4.62lb/*7.71lb	3.37lb restricted
Index to thumb pinch force	2.75lb/*4.63lb	7.64lb
Power grip	22.48lb/*30.64lb	16.86lb
Load limits per digit	70.55lb	32.27lb
Load limits overall	198.42lb	70.55lb
Push up from w/c, hand closed	183lb	198.42lb
Push up from w/c, single digit	28.5lb	13.23lb
Carry heavy bag, full hand	231lb, no fail	?
Carry heavy bag, one digit	103lb, no fail	?
Carry heavy bag, thumb	79lb	?
Weight, small / medium hand	1.014lb	1.18lb
Weight, regular / large hand	1.025lb	1.19lb

*with pulsing

BATTERY OPTIONS

The i-LIMB Pulse has two recommended internal battery options to choose from for best performance. The 2400 mAh capacity battery has a charge time of 6-7 hours while the 1300mAh battery has a charge time of 3 hours. The bebionic v2 hand also has two recommended internal battery options to optimize performance. The Single Battery(BBI=2200) has a 2200mAh capacity which takes 3.5hrs to fully charge from a depleted state. The Split Cell Battery(BBI=1300) has a 1300mAh capacity which only takes 2 hrs to fully charge from a depleted state.

Both systems, RSLSteeper and Touch Bionics, Inc. recommend charging the battery every night, regardless of usage. When considering the installation of a wrist rotator the recommendation should lean toward use of the 2200mAh(RSLSteeper)or 2400mAh(Touch Bionics, Inc.) battery options.

These battery options also help support consistent communication (blue tooth or RF module/USB dongle) with the hand during programming.

GLOVE OPTIONS AND SIZING

The i-LIMB Pulse has 3 covering options which can be utilized to match the patient needs. The i-LIMB skin offers a minimalistic covering approach which matches the mechanical contours and details of the hand, available in 4 color options and 2 hand sizes. The i-LIMB High Definition Covering offers the patient a more cosmetically appealing and durable cover, available in 10 color options, male or female and 2 hand sizes. The i-LIMB Pulse has a third option for covering, which is a Custom High Definition Covering. This option provides the patient the most realistic match to the contralateral hand and arm size, shape and coloring details. The i-LIMB Pulse is available in two specified sizing options, Regular and Small.

The bebionic v2 hand currently has one covering option available for each hand size. This covering system, the bebionic glove, is a variable hardness multilayered glove, mesh lined, available in 20 colors, fitted with custom made silicone factory fitted nails and integral silicone thimbles at the digit tips for additional grip compliance. The bebionic v2 hand is available in two sizes, Large and Medium.

DISCUSSION

We have taken an objective viewpoint from our experiences to date. Each hand represents an advancement in functional positioning for the user. As new multi-articulating hands enter the market, it will become increasingly important for the clinical team to understand the capabilities of each hand. This comparison creates an up to date way of seeing the prosthetic hand features including available grip patterns, functional programming, graphic user interface, component compatibility, available load testing/grip force summary, battery options, glove options and sizing.

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DESIGN OF A HYDRAULIC HAND PROSTHESIS, WITH ARTICULATING FINGERS

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INTRODUCTION

Rejection rates of body-powered hands are higher than that of hooks.[1] Body powered hands are inefficient. As a result they require an uncomfortable high activation force, and produce a relatively low pinch force in return (<15 N).[2] Also they have stiff fingers, which do not adapt to the shape of the grasped object. Despite all the drawbacks of the current hands, the design of body powered hand prostheses almost has not changed since the 1950's. The activation force has not been reduced. The pinch force is still low, and hand prostheses are still quite heavy. There have been attempts to increase the efficiency of body powered hand prostheses, by using hydraulics.[3, 4] However, these studies have not resulted in the commercial application of hydraulics in body powered arm prostheses.

GOAL

The goal of this study was to design a new body-powered, voluntary closing, hand prosthesis, which has articulating fingers. This hand should require an operation force within a comfortable level and should have a low mass.

METHODS

An articulating voluntary closing hand was designed and prototyped. Before the hand was designed, a number of demands and boundary conditions were defined.

Hydraulics

The principle of hydraulics was used, to transfer the energy of the body movements to the fingers of the hand prosthesis. Using hydraulic offers some potential benefits. In the first place, using hydraulics makes it possible to abandon the use of an Bowden-cable. This can significantly improve the efficiency of the entire system.[3] The Bowden-cable dissipates a significant amount of energy, due to the friction in the curves of the cable. The curvature of a hydraulic hose has no significant influence on the efficiency of the system. In the second place, the use of hydraulics makes the energy distribution amongst the individual joints and fingers easier and more efficient.

The hydraulic actuators should fit inside a finger, to enable actuation of the Proximal Interphalangeal (PIP)-joint. An actuator should weigh less than 10 grams, to stay within the overall mass limit of the entire hand. The actuators should be able to operate a pressure level of 50 bar, to enable a high maximum pinch force. Currently there are no standard hydraulic components available, which meet such strict requirements. Therefore miniature lightweight cylinders were designed, for the hydraulic hand prototype. Water will be used as a hydraulic medium, instead of hydraulic oil, to reduce the negative effects in case of a small leakage.

Boundary conditions

- It was decided to design a hand of with a size of 7 ¾. This corresponds to a small size adult male hand, or a large size female hand. Once a prototype has been build, it can be slightly expanded or reduced to create a larger or a smaller size.
- The hand prototype should be suitable for body powered shoulder control.
- The elements of the hand should be modular, to enable easy replacement and upgrading of subsystems of the prototype.
- Three fingers should have at least two actuated Degrees of Freedom (DoF's) each, the Metacarpophalangeal (MCP)-joint and the PIP-joint. The little finger should have at least one actuator and two actuated joints. In total the hand should have at least seven actuated degrees of freedom.
- The thumb should have at least one passive controllable DoF. However, by making use of modularity, the thumb should be easily be replaceable by an actuated thumb.
- The wrist should have at least two passive DoF's. One which enables for pro- and supination. A second which enables for flexion and extension of the hand.

Under-actuation

The multiple slave actuators in the hydraulic hand, should all be controlled by one master cylinder. Therefore the control will be done by using the principle of under-actuation. A system is by definition under-actuated when

there are more DoF's than controlled actuators.[5] As a result the configuration of the fingers dependents on the external forces acting on them.

Demands

The most important demands to which the hand should comply are:

- The mass of the hand should be as low as possible. The hand should weigh significantly less than current available hands (which weigh around 350 gram). The goal was to design a hand which has a mass below 100 gram.
- The activation force should be at a comfortable level. The comfortable activation force is not exactly known from literature. There are indications that the maximum comfortable level is around 50 N.[2]
- The maximum pinch force should be above 30 N, to enable a broad range of activities of daily living.[6]

The hand should be very efficient, to enable a high pinch force, at a low activation force. Therefore the hand should be designed to have a very low amount of hysteresis.

RESULTS

A hydraulic hand was designed and constructed. The hand has 7 DoF's, actuated by 7 hydraulic cylinders. Three fingers have 2 actuated DoF's, the little finger has one actuated DoF. The thumb has one passive DoF (Figure1).

Figure 1: The CAD-design of the hydraulic hand

The predicted mass of the hand was less than 100 grams, according to the CAD-model. The measured mass of the ungloved hand prototype was 110 grams.

The hand fits inside a cosmetic glove of size 7 ¾. The fingers are actuated by miniature hydraulic cylinders. The proximal cylinders in the finger have a piston diameter of 8 mm, the distal cylinders and the cylinders in the little finger have a piston diameter of 7 mm (Figure 2).

Initial testing showed that the cylinders could be operated at a pressure exceeding 50 bar, without any problem. Initial pinch force measurements showed that a finger could produce a pinch force of more than 30 N.

Figure 2: One of the three 2DoF hydraulic fingers. The finger is activated by the small metal cylinders.

DISCUSSION

The mass was 10 grams higher than predicted. This is caused by the fact that some parts were not included in the CAD-drawings. The mass of the hand can be reduced, by further optimisation of the hand frame. The frame was not optimised for a low mass in the current prototype.

The measured pinch force of more than 30 N, complies with the demands. Further measurements should determine the maximum pinch force. The pinch force is limited by the maximum allowable system pressure. Also the activation forces should be measured, which are required to pinch at certain pinch force levels. The required force should be within the comfortable activation force. The transmission ratio for the activation

Future work

The final goal of the project is to have the hydraulic hand clinically tested. Before the hand can be clinically tested the following steps have to be executed:

- The hand will be mechanically tested. The goal of these tests is to determine the maximum pinch force,

the grip strength, the required activation force, and the mechanical efficiency.

- The hand will be subjected to a durability test, to determine how long the hand can operate without failure. The hand will be adapted, when necessary.
- A special master cylinder will be designed, to enable body powered shoulder control.
- Initial clinical test will be performed with healthy subjects, by means of a prosthesis simulator.

CONCLUSIONS

Current body powered hands are inefficient, and do not have articulating fingers. They have a low pinch force (<15 N) and require a high activation force. Therefore a new hand was designed and prototyped. The hand has the following specifications:

- The ungloved hand has a low mass of only 110 gram.
- The hand is controlled by the principle of under-actuation. One master cylinder controls 7 slave hydraulic cylinders.
- Three fingers have two cylinder actuators each. The little finger has only one actuator.
- The custom designed miniature hydraulic cylinders have a diameter of 8 and 9 mm. The cylinders fit inside a finger of a cosmetic glove.
- The cylinders can be operated at a high pressure (>50 bar).
- Initial measurements show that the hand can pinch over 30 N.
- Further testing is required to determine the required actuation force. The required activation force can be optimized by optimizing the transmission ratio.

ACKNOWLEDGEMENTS

We would like to acknowledge John Dukker, for fabricating the parts and for his technical assistance.

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CASE STUDY: MULTIPLE-LIMB AMPUTEES FIT WITH POWERED PARTIAL HAND PROSTHESES

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INTRODUCTION

For people with digit amputations, prosthetic choices have been limited to cosmetic restorations usually made of silicone, opposition posts, or mechanical linkages. In a study in Australia, those with partial hand losses perceive themselves to be at a higher disability than those with unilateral transradial or transhumeral upper extremity amputations.[1] More than half of partial hand amputees are unable to return to their previous work.[2] Of those that did return to work, the majority did not find their prosthesis functional for work.[2] Articulating partial hand prostheses have been available since at least the mid 1970's, but have not largely been adopted due to limitations of fit or usefulness.[3,4]

Each year approximately 17,000 digit amputations are performed in the United States.[5] Between 2000 and 2010 it is estimated that within the Veterans' Administration there were 3000 digit amputations. There are roughly 400 warriors with digit amputations attributable to operation OIF/OEF, and an estimated 50 with bilateral thumb loss.

Improvement in the ability to carry out activities of daily living (ADL) using partial hand prostheses is evaluated in two case studies of subjects presenting with multiple-limb amputations. Impairment and disability are assessed using the *quick*DASH, the Jebsen-Taylor Hand Function Test, and the Box and Blocks Test during initial visits and at follow-up visits after delivery of their prosthetic devices. Both of the subjects represent unique challenges for prosthetic fitting and have revealed unique outcomes in their abilities to utilize their partial hand prosthesis in ADLs.

CASE STUDY 1

Subject Information

Subject is a 33-year-old male Marine who suffered multiple wounds due to an improvised explosive device (IED) blast in September 2010 while stationed in Iraq as part of Operation Iraqi Freedom (OIF). He was treated in Al Asad and then transported to Balad Military Medical Base before being admitted to the James A. Haley Veterans' Administration (JAHVA) Polytrauma Unit in October 2010. His injuries include traumatic brain injury (TBI), right transradial amputation, and left partial hand amputation where the 1st digit was partially amputated, and digits 2, 3, 4 were completely amputated (Fig. 1). Prosthetic care was provided by the VA while the subject was an inpatient.

Figure 1: Subject 1 shown with right transradial prosthesis and left partial hand amputation.

Prosthetic History

The subject was initially fit for a right body-powered prosthesis constructed as a traditional hard laminate socket with a figure-8 harness, triceps cuff, flexible hinges and hook terminal device (TD). The subject quickly mastered the use of this prosthesis. In December 2010 he was provided with dual site externally-powered myoelectric prosthesis with an I-Limb (Touch Bionics) terminal device. Independent donning/doffing was not achieved by the subject due to the limitations of his left hand. Clinic reports indicate that the subject stated he was pleased with the myoelectric prosthesis, but he often came to appointments without the arm in place. The subject experienced limb volume loss resulting in difficulty maintaining sufficient contact to operate the controls effectively. Eventually, new sockets were made for both prostheses. The subject was provided with a custom silicone cosmetic cover for his myoelectric prosthesis. At the end of 2010 the decision was made for JAHVA to provide him with a partial hand prosthesis on his left side.

Treatment

At the time of the fitting for the left partial hand prosthesis, the subject was an inpatient at the JAHVA, and was 6 months post-injury. The limited thumb digit had a significant impact on his ability to grasp. Because of the bilateral involvement, it was deemed especially important to reestablish grasping functions. For this reason, we elected to provide a prosthesis

incorporating prosthetic digits (ProDigits) manufactured by Touch Bionics. ProDigits are newly available self-contained prosthetic digits that are individually powered and controlled to provide new fingers for partial hand patients. This is the initial experience JAHVA has using ProDigits. Individual devices can be configured to match the number of digits required for partial hand restoration.

Subject was cast for a partial hand prosthesis using silicone which was sent to Touch Bionics for fabrication. A flexible silicone socket resembling a sleeve with a carbon fiber frame was fabricated (Fig. 2). The sleeve extended from distal to the bicep cubital fold to the distal end of his residuum. A zipper was incorporated to enable donning and doffing. The carbon fiber frame fit over his residuum, distal to the wrist, leaving the thumb exposed. Electrodes were placed over the wrist flexors and extensors. EMG sites were also identified on the residual hand; however, the decision was made to utilize forearm sites to use consistent controls between the contralateral transradial prosthesis and the partial hand prosthesis. The prosthesis incorporated three ProDigit fingers (to replace digits 2-4). Because this was the first prosthesis provided by this hospital that integrated ProDigits, Touch Bionics flew in a certified prosthetist and occupational therapist to assist with fittings. Several modifications were made to the socket in order for the subject to independently don and doff the partial hand prosthesis by using his right transradial prosthesis. The first modification was the addition of a ring on the posterior zipper, this allowed him to hook the thumb of the I-Limb into the ring to zip or unzip. The second modification was relocating the positioning of the on/off switch in order for him to activate it without difficulty.

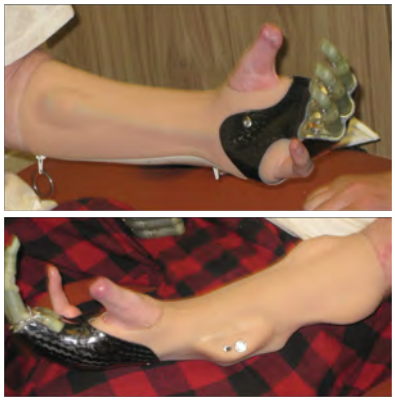

Figure 2: Partial hand prosthesis with ProDigits.

Specific activities that had presented a problem for this subject included activities of daily living (ADLs) such as cutting up food and toileting. During therapy independence in those activities was achieved (Fig. 3). Initially, Touch Bionics delivered the prosthesis without the addition of a thumb post. During their training session with the occupational therapist they deemed it beneficial for the subject to receive a digital restoration of the partial thumb to improve grasping. Subsequent testing was not performed with the thumb post in place; that addition came later. Results from the three tests are provided in Outcomes section of this paper.

Figure 3: Subject 1 performing a simulated meat cutting exercise during therapy shortly after delivery of prosthesis.

CASE STUDY 2

Subject Information

Subject 2 is a 50-year-old male Navy veteran who had bilateral transtibial amputations and a right partial hand amputation (Fig. 4) involving digits 2-5 secondary to heparin-induced thrombocytopenia following a massive myocardial infarction (MI) which occurred in March 2005. 2005, Two months after the MI, he underwent a heart transplant, requiring him to take immunosuppressant drugs which now cause hand tremors. Complicating prosthetic use, he suffers from right shoulder restricted range of motion (ROM) with pain at end range, but has functional ROM in his right wrist.

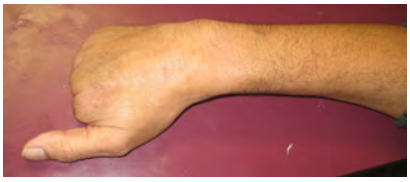

Figure 4: Subject 2 shown with right partial hand amputation.

The subject has recovered remarkably well from the heart transplant and is ambulatory with his lower extremity prostheses, using a cane when walking for extended periods of time. His most current goal is to be fit with an upper extremity partial hand prosthesis that will provide him with a functional grasp to aid in activities of daily living (ADL). He learned to use his thumb and palm very effectively when grasping or picking up certain objects, but most activities involving hand function are limited. The subject seeks to regain the ability to perform a wider range of tasks that involve the use of his hands.

Prosthetic History

Two months following his surgeries, the subject was fit with bilateral transtibial prostheses by the JAHVA; he later chose an outside provider for future prosthetic needs. He uses a pin suspension system and is ambulatory, using a cane for extensive periods of walking. He is able to independently don/doff his prostheses. The partial amputation of his right hand was followed by an extended period of sensitivity in the residuum which delayed prosthetic fitting. He learned to utilize his thumb and palm when performing ADLs, even so, he is severely limited in functional capabilities involving the hands. In early 2009, the sensitivity in his right hand improved and he was ready for prosthetic fitting. Commercially available options for functional prostheses that would fulfill his needs were scarce. He was fit with a passive custom silicone partial hand prosthesis. Later that same year he requested and was provided with a recreational upper extremity prosthesis which could be used to play golf. In 2010, the subject was fit with a body-powered partial hand prosthesis that utilized four mechanical fingers operated via wrist flexion. This device was eventually rejected by the subject. In 2011, JAHVA provided the subject with an externally-powered myoelectric prosthesis equipped with ProDigits developed by Touch Bionics. See figure 5

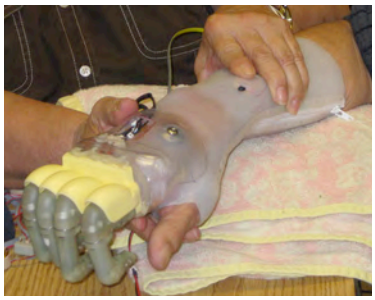

Figure 5: Subject two wearing ProDigits

Treatment

This was the second partial hand prosthesis provided to a veteran by the JAHVA that incorporated ProDigits. Subject's residual limb was cast using silicone; the mold was shipped to Touch Bionics for prosthetic fabrication. A flexible silicone socket (similar to a sleeve) with a carbon fiber frame was fabricated. The flexibility of the socket accommodates the natural movement of the skeletal and musculotendinous structures and allows for wrist motion in all planes; this characteristic makes it possible to achieve maximum function with a prosthesis. The intimate fit helps achieve maximum suspension and facilitates maintaining good electrode to skin contact over myoelectric sites. The flexible portion of the socket extended from approximately the bicep cubital fold to the distal end of his residuum. The subject chose black silicone as opposed to something more flesh-colored for the flexible socket. A zipper was incorporated to facilitate donning and doffing. The carbon fiber frame fit over his residuum, distal to the wrist, leaving the thumb exposed. Electrodes were placed over the muscles controlling wrist ulnar and radial deviation; this helped preserve functional wrist flexion and extension. The prosthesis included four ProDigit fingers to take the place of missing digits 2-5. Open/close controls were set to operate with ulnar/radial deviation. Electrode gains required adjustment over time because his tremors would elicit inadvertent movements of the ProDigits during training.

OUTCOME MEASURES

Three measures were used to follow the subjects; the Box and Blocks Test, The Jebsen-Taylor Test of Hand Function, and the Disability of Arm, Shoulder and Hand Assessment (*Quick*DASH). The box and blocks is a test of manual dexterity. The test was originally developed to evaluate adults with cerebral palsy. The Jebsen-Taylor is a seven-part test which evaluates a broad range of everyday hand functions using common items such as paper clips, cans, pencils, etc. The *Quick*DASH is a shortened version of the DASH Outcome Measure, which uses 11 items to measure physical function and symptoms in people with any or multiple musculoskeletal disorders of the upper limb. Measurements were scheduled to be taken at three time points; with no partial hand prosthesis during the casting visit, after delivery and two days of occupational therapy with the new prosthesis, and 45 days post-delivery. Tables 1 & 2 provide test results for Subjects 1 and 2, respectively.

DISCUSSION

*quick*DASH

The *quick*DASH is a survey, and is reflective of the user's perceptions. The *quick*DASH score for Subject 1 indicates improvement between visit one and two subject 1. For comparison to Davidson's study, prior to provision of a prosthesis, Subject 1 has a score that falls between Bilateral Upper Extremity amputations (68 ±5) and Partial Hand Amputation (49 ±22). [1] His second score, after two days of occupational therapy post-delivery, reflects a large improvement; a lower score is better. After 45 days, the score shows a loss in improvement.

Subject 2 was not tested at 45 day post-delivery for unrelated medical reasons which delayed follow-up. He was tested while wearing his body-powered partial hand prosthesis that he rejected. The results show a loss in function with the body-powered, and no change between no prosthesis and the ProDigits at post-delivery visit 2. The subject has attend therapy sessions since visit 2 and reports that he is very pleased with the ProDigit prosthesis, stating he wears

the device 4-5 hours per day. He comes to visits wearing the prosthesis.

Table 1: Outcome Measures for Subject 1 at 3 time points.

Test	Subject 1 Outcome Measures		
	No Prosthesis	***2 Days Post Delivery***	***45 Days Post Delivery***
quickDASH (Score)	59	34	41
Box & Blocks (# Blocks)	35	8	22
Jebsen-Taylor	(time/fraction completed)		
• Writing	1:04 – 20/24	2:00 – 0/24	:48 – 24/24
• Turning Cards	0:07 – 5/5	0:22 – 4/5	0:23 – 5/5
• Small Objects	0:20 – 5/5	0:48 – 4/5	1:41 – 2/6
• Feeding	0:15 – 4/4	1:02 – 5/5	0:30 – 5/5
• Stacking Checkers	0:27 – 4/4	0:56 – 4/4	0:40 – 4/4
• Light Cans	0:04 – 5/5	0:40 – 5/5	0:45 – 5/5
• Heavy Cans	0:04 – 5/5	0:33 – 5/5	0:20 – 5/5

Table 2: Outcome Measures for Subject 2 at 3 time points.

Test	Subject 2 Outcome Measures		
	No Prosthesis	***Post Delivery Body Powered***	***2 Days Post Delivery (ProDigit)***
quickDASH (Score)	41	47	40
Box & Blocks (# Blocks)	37	16	29
Jebsen-Taylor	(time/fraction completed)		
• Writing	0:38 – 24/24	0:35– 24/24	:38 – 24/24
• Turning Cards	0:18– 5/5	0:48 – 5/5	0:18 – 5/5
• Small Objects	0:14 – 5/5	1:41 – 4/5	0:36 – 6/6
• Feeding	0:13 – 4/4	1:40 – 5/5	1:14 – 5/5
• Stacking Checkers	0:09 – 4/4	0:43 – 4/4	0:19 – 4/4
• Light Cans	0:05 – 5/5	0:18 – 5/5	0:21 – 5/5
• Heavy Cans	0:05 – 5/5	0:23 – 5/5	0:32 – 5/5

Box and Blocks Test and Jebsen-Taylor Hand Function Test

The Box and Blocks Test and Jebsen Taylor Hand Function Test are skill based tests. Dromerick *et al*. showed that immediately upon receipt of a prosthesis, function actually goes down.[6] The team indeed saw that also with both subjects. This may have been further exacerbated because this was a new prosthetic user with a traumatic brain injury. Secondly, we may be approaching a ceiling effect on some measures. According to the results of Hackel et al.,[7] this subject is approaching normative values for the light cans, heavy cans, and card turning tasks. Similarly the starting score on the box and blocks is relatively high. The partial thumb digit also played a role in the functional challenges during visit two. With the finger restoration, his thumb was now short by comparison, and he had more difficulty grasping with the prosthesis, than by using the thumb without a prosthesis. A decision was made to provide a passive restoration to the thumb to improve opposition.

Subject 2 has become very adept at manipulating objects such as those used in the Box and Blocks and Jebsen-Taylor without a prosthesis. The initial testing after delivery of his prosthesis indicates that he has to slow down to grasp and move the objects. His ability to perform actual day-to-day functions has improved, according to self-reporting. This important information is not reflected in the chosen tests.

CONCLUSION

New partial hand restoration for a two complex cases involving multiple-limb amputees was presented; one was further complicated by the presence of traumatic brain injury. The ProDigit is a promising device for the treatment of the partial hand amputee. In each case, the ProDigits prosthesis was very well received by the users. The functional testing for Subject 1 did not reflect the increase in functionality that the user described or articulated in the *quick*DASH instrument. In the case of Subject 2, the initial *quick*DASH score did not reflect what was observed (he successfully utilized the device). Hopefully, with continued follow up and monitoring, functional testing will reflect the users' perceptions. At least for these users, the functional tests selected (Box and Blocks Test and Jebsen Taylor Hand Function Test) may not be sensitive to functional changes related provision of a partial hand prosthesis.

ACKNOWLEDGEMENTS

This material is the result of work supported with resources and the use of facilities at the James A. Haley Veterans' Hospital, the Tampa Regional Amputation Center, and Prosthetics and Sensory Aids Service.

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THE HIGH-FIDELITY INTERFACE: SKELETAL STABILIZATION THROUGH ALTERNATING SOFT TISSUE COMPRESSION AND RELEASE

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ABSTRACT

Traditional interface designs have largely focused on tissue containment of the encapsulated limb and establishing stability via anatomical contouring in the areas of the interface closest to the proximal joint of said limb. Firm control of the shaft of the underlying bone of the encapsulated limb has either been wholly ignored or given only a cursory examination at best. Indeed, for many there still remains a question whether or not the underlying bone can be controlled at all. Limited biomechanical knowledge, a general acceptance of Hydrostatic theory with regard to interface design, a glaring absence in our quantification of window edema and its relation to aperture design and location and the relaxed pace at which we have both developed an interest as well as the associated technology to assess the socket environment in a comprehensive fashion have all inhibited rapid interface advancement. A new theory "Compression-Release Stabilization" (CRS) focuses primarily on control of the underlying bone and forms the foundation of the High-Fidelity Interface (HiFi) described herein. This paper will discuss the theory as well as the results of its application in both upper and lower limb prosthetics and orthotics.

INTRODUCTION

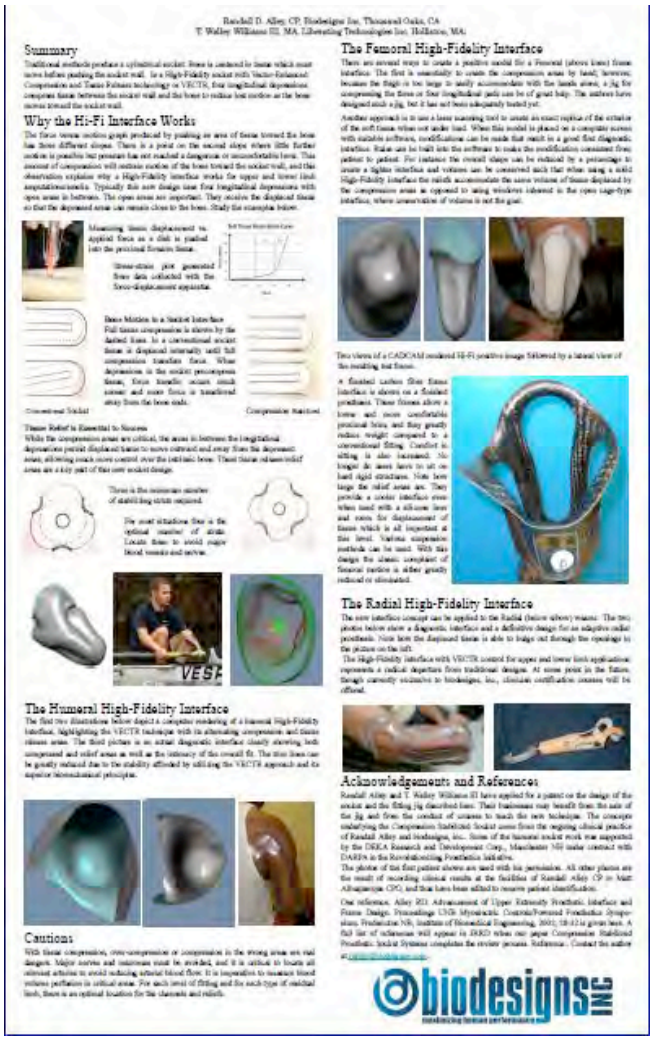
Newton's First law essentially states that an object at rest will remain at rest unless acted upon by sufficient force to create a change in state. With regard to even the latest interface designs from the Symes level to the hip and partial hand to the shoulder, in nearly every case, the underlying bone or bony structures translate significantly within the interface relative to the interfacial boundary during volitional movement and in response to externally applied loads. This unwanted motion is predominantly due to the lack of sufficient counterforce generated by the soft tissue between the moving bone and the socket wall. Because the underlying bone is typically fixed at one end, it swings in such a way as to cause its distal end to strike the interface wall, separated by a very thin layer of highly compressed tissue. In this paper I illustrate the inherent design weakness of traditional sockets [1,2] and why a different model based on alternating soft tissue compression and release applied along the shaft of the underlying bone or strategically about targeted bony structures offers a more efficient way to generate prosthetic motion. Vastly improved stability, enhanced functional range of motion, improved ability to handle (position and carry) or ambulate with greater loads more comfortably, reduced energy expenditure, increased gait speed and stride length, and a perception of the prosthesis feeling more like a part of the wearer were all achieved. Its intimate connection with the limb offers the wearer a feeling of agility and precision that, based on the laws of physics, cannot be equalled with a traditional approach. Finally the patient subjects reported a perception their prosthesis weighed less than their traditional system, even in instances where the prosthesis employing Compression-Release Stabilization weighed more than its traditional counterpart.

The alternating soft tissue compression and release technique places longitudinal compression areas along nearly the entire shaft of the bone or underlying bony structures while the release areas allow for soft tissue, including skeletal muscle, to escape out of the fields of compression. In many cases, primarily in upper limb, this outward flow of tissue is completely unrestricted to allow for increased heat dissipation but also to reduce the overall volume of tissue that lies within the field of compression. With correct aperture design to control a variety of critical variables including the volume of released tissue, the rate of "step-off" from elevated compression to zero compression, exiting skin tension, fluid and venous return, as well as the strategic location of the release areas themselves, window edema is not a concern. Likewise, correctly applied compression to facilitate bone capture and control, if designed and deployed with precision regarding shape, location, extent and compression level, blood perfusion is also not a concern.

Because less tissue remains between the compressed area and the shaft of the bone, additional compression-and therefore skeletal control-is gained. In essence, by "precompressing" the overlying soft tissue prior to volitional movement or applied external force, its density is significantly increased. Subsequently, as the compressed tissue's density increases, so too does its ability-when trapped between the

target bone and the interface wall-to provide a counterforce to unwanted skeletal motion [3].

Because it can be understood that preloaded or highly compressed tissue within the interface will not only provide a greater counterforce but will provide it more rapidly, we can apply Newton's Law to assume the prosthetic interface will respond more quickly as well. If we then consider the total area of elevated compression at the boundary of bone to soft tissue and relate it to Archimedes' Principle regarding buoyancy, we can analogize that with any increase in surface area of the compressed region at the bone to tissue boundary (the hull) there is a corresponding decrease in the force (buoyancy) required to initiate prosthetic motion. Thus if we increase both the magnitude and total area of increased density at the bone to tissue boundary, we can readily appreciate the improved biomechanical condition existing within the interface.



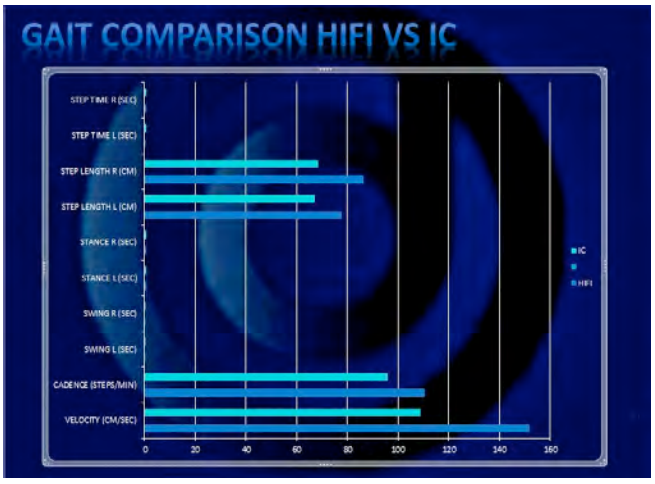
METHOD

A series of assessment tools are currently being developed in anticipation of formal clinical trials in several locations throughout the United States and Europe. The results of informal analysis provided here attempt to simply illustrate trends observed in both upper and lower limb applications.

Close to one hundred patients have been fit with the High-Fidelity Interface to date and at the time of this writing.

For the purposes of this preliminary study, emphasis of this paper will focus upon a detailed discussion of the prosthetic application of CRS, including both measured and observed results. A heart rate monitor was used over a fixed distance to assess energy expenditure. The theoretical basis of this measure is the linear relationship that exists between heart rate and energy expenditure (EE) in steady state exercise involving large muscle groups. The method has shown to have high reproducibility within subjects [4]. Distance was measured at the conclusion of a two minute walk test comparing traditional ischial containment to the HiFi interface. A randomized crossover study is proposed in the future at several sites to validate the acquired data.

Gait attributes were assessed utilizing a GAITRite mat [5]. Video assessment of upper and lower limb wearers was also undertaken along with a patient survey regarding the High-Fidelity Femoral Interface and its wearers' subjective perceptions.



RESULTS

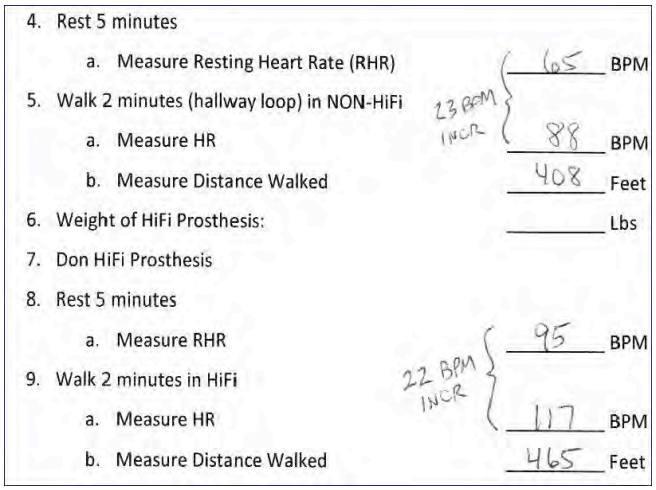
The above table demonstrates a sample data set comparing a traditional ischial containment socket to the High-Fidelity Interface, averaged over three walks on a GAITRite mat for a single individual. The clinical protocol as a condition for licensing this design involves informal analysis of gait

parameters and HR, and so much larger patient populations will be the subject of future papers.

Videos to be presented illustrating range of motion of an upper limb wearer involved in the Luke Arm Project under significant load show the increase in both range of motion and comfort.

Finally, patient satisfaction surveys regarding their experiences with the High-Fidelity Interface reveal a greater level of satisfaction with the newer design.

The sample below shows the HR delta between rest and post-exercise to be relatively equal, though the subject in the HiFi walked 57 feet farther in the same two minute interval. It is interesting to note the resting HR prior to the subject's second walk test was significantly higher than his original resting HR.



DISCUSSION

The gait data illustrate the potential benefits of Compression-Release Stabilization, and although merely a snapshot of the results, significant data with much larger upper and lower limb patient data sets are planned for the future. The difference between a traditional ischial containment and the HiFi interface with regard to the delta between rest and post-exercise measurement was approximately 4% in favor of the HiFi, while the difference in distance walked given the same time period amounted to a significant increase of approximately 22% for the HiFi.

In no way do these singular cases allow us to draw formative conclusions as to expected results for all patients, but given the simple biomechanical nature of Compression-Release Stabilization and the laws of nature, it is fairly easy to understand why it might improve function for both upper and lower limb wearers in significant ways.

CONCLUDING REMARKS

With regard to upper limb, studies involving range of motion analysis under light and heavy loads, positional accuracy without visual aid, stability and bone motion within the interface are among some of the areas of interest. In lower limb, energy expenditure, preferred gait velocity, step length and overall gait quality will be assessed. In both upper and lower limb cases, heat dissipation and other temperature-related characteristics will be studied. Finally, a quality of life survey will be given to both upper and lower limb wearers of the High-Fidelity interface to assess their subjective impressions.

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CONTROLLING TWO INDEPENDENT JOINT MOTIONS WITH THE ACROMION

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INTRODUCTION

Recently two groups reported excellent results having subjects control two degrees of freedom (DOF) by sampling the motion of the acromion. They use a cap placed over the acromion to move a joy stick. Since joy sticks are difficult to use clinically, we built an X-frame socket to use as a test bed to try other approaches to recording this motion. This test socket can use cables to activate a linear transducer for continuously recording the motion of protraction-retraction independent of motion in elevation-depression. A second transducer can record changes in elevation. By using cables, the inputs can be separated better than with a joy stick, and furthermore better feedback can be provided to the user.

A socket interface for free motion of the acromion

Traditional X-frame sockets encapsulate the entire lateral aspect of the remaining shoulder. This constraint makes the location of stable myoelectrode sites easy, but it severely limits independent motion of the acromion. Any attempt to move the tip of the shoulder forward results in forward rotation of the entire socket. To capture this motion prosthetists have used elastic webbing in the cross-back harness. Typically a user protracting the shoulder will cause a rotation of the socket with respect to the contralateral side of 15 to 30mm at the acromion. Compare this to the motion of the free acromion with respect to the thorax as reported by Williams and Lipschutz. [1, 2] They measure two to three times as much displacement. Losier et al also report using the acromion to control two degrees of freedom. [3]

Designing a socket where the acromion is free

The author had colleagues make a series of photos without a socket followed by another series with a socket designed for study. (Note while viewing these photos that my right shoulder is lower than the left when relaxed.) Photos were made from directly ahead of a white board with a horizontal line on it, and the camera position was kept the same throughout. In Figure 1, a mark has been placed over the sternoclavicular joint with a second over the acromioclavicular joint. The clavicle is also outlined. A careful measurement was made between the tips of my glasses and this measurement was used to quantify changes when I moved. In figure 1, the mark on the acromion moves up 17mm and medially 6mm. The angular motion of the clavicle is 19.4° which is no surprise, since the tip of the acromion is constrained to move about the center of the sternoclavicular joint. In the test bed socket elevation-depression will be motion in a plane tilted 19.4° from the vertical. Figures 1 and 2 show relaxed to max elevation.

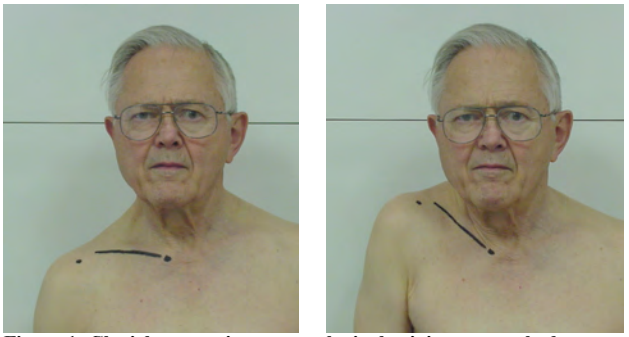

Figure 1: Clavicle, acromion, sternoclavicular joint are marked

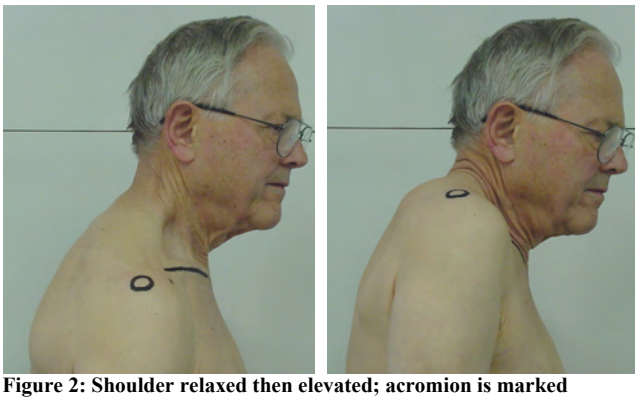

Figure 2: Shoulder relaxed then elevated; acromion is marked

To optimize the socket you need to know which parts of the anatomy move and which do not. For this we marked areas which showed no motion to palpation when the acromion was moved to all four maximum displacements. Figure 3 shows the result. Unfortunately the frame in Figure 4 was trimmed out before these marks were made or it would have covered more of the area in the back.

Lessons learned from the test frame

The shoulder cap in Figure 4 was made directly on the subject using a low-melting-point plastic. (The first cap, made from the cast for the frame, was too loose.) To make the cap a separate cast should be taken with the prosthetist pushing down around the area which will be the edge of the cap as the plaster sets. Note the four elastic bands holding

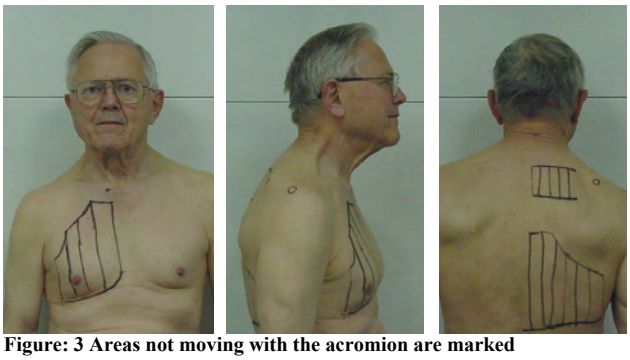

Figure: 3 Areas not moving with the acromion are marked

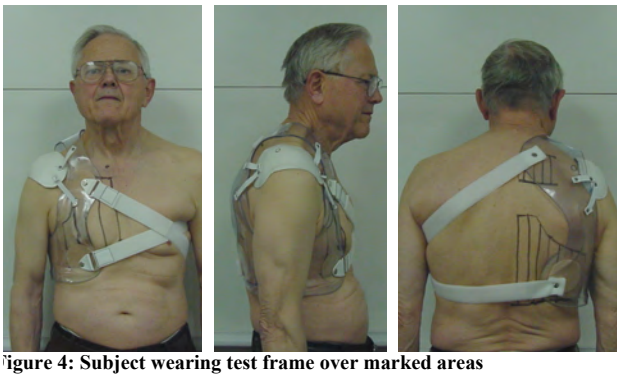

Figure 4: Subject wearing test frame over marked areas

the cap against the subject. Their angular location, length, and pre-stretch determine the force vectors holding the cap. These forces must remain active as the subject positions the shoulder tip. In addition the bands must not interfere with the control mechanisms added later. Typically in a finished prosthesis the elastics would exit from between two thin layers comprising the cap.

Where should we measure motion?

Lipschutz et al measured motion of the acromion with a joy stick at the highest point on their test frame near the user's neck. Attached to the stick was a rod passing through a ring on the shoulder cap to accommodate the change in distance with elevation and depression. The joy stick had two potentiometers, one recording forward-back angular motion and a second recording up-down. This arrangement produced good data, but having a joy stick here is too uncosmetic in a definitive prosthesis. This paper explores an alternate scheme for collecting the same data.

A simple cord pulling a linear transducer is good for detecting changes in the distance between two points. We analyzed over 35 photos which showed linear displacements of about 100mm. The LTI Linear Transducer has too short a range to record this, since it can only detect motions of 0-12 or 0-25mm. Thus a different type of transducer is needed. A good solution to would be a thin capstan about 35mm in diameter above a thin coaxial potentiometer. These parts can fit within a disc 40mm in diameter and only 6mm thick. The cord goes around the capstan which has a spring to maintain some tension in the cord. To record protraction-retraction, the cord needs to pass from the cap medially almost to the center of the back where a small pulley can redirect the cord to a convenient transducer mount under the scapula.

While protraction-retraction seems to move the acromion forward and back, the actual motion is more complex. Study Figure 5 which is a horizontal cross section through the frame at the height of the sternoclavicular joint with the cap in its neutral position. The joint center is marked with a black dot in the upper right. A second mark has been placed at the other end of the clavicle where it is constrained to move on the magenta arc. The plastic frame is indicated by heavy black lines and the cap by a thin blue arc. Two possible cord locations are indicated. The location on the subject's back is on the left in green and the front location is on the right in brown. A black bar shows how far the center of the cap moves left-right when photographing the frame from the side. The cord on the back is almost tangent to the arcuate motion of the cap which will result in maximal motion, while the front cord is almost radial and parallel to the dashed line resulting in little motion. Thus we need only place a transducer and cord in the back.

To locate the pulley attachment, we studied the photos to locate the center of rotation during elevation-depression. It is below the upper harness-strap rivet in Figure 4 and a little below the frame edge. The frame needs to be larger so a small pulley can be placed here. This location minimizes cross talk between the front-back and up-down motions.

To record elevation-depression a cord can be run from a fixed point below the cap in front and then over the cap in a Bowden sheath to a second transducer in back.

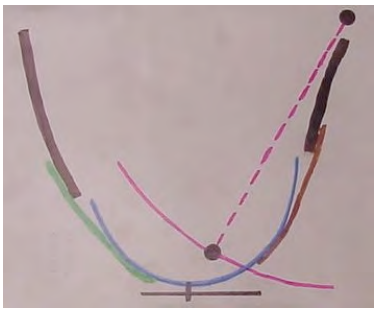

Figure 5: Horizontal plane at height of the sternoclavicular joint

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DEVELOPMENT OF AN INTRAOSSEOUS TRANSCUTANEOUS AMPUTATION PROSTHESES (ITAP)

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INTRODUCTION

Using artificial limbs to restore function for amputees is a concept that dates back to approximately 700BC. Despite this, attaching artificial limbs to the body remains clinically challenging, with inadequate mechanical fixation, poor stump-socket fitting particularly to short residual limbs, friction leading to the development of pressure sores, infection of the stump soft tissues and sweating often leading to limb disuse [1-3]. Eliminating the socket, by directly attaching the artificial limb to the residual bone through osseointegration transmits forces through the bony skeleton alleviating the problems associated with the socket. This technology also has the potential to increase the range of motion of the proximal residual joint particularly with humeral amputees, permits comfortable sitting, and has been shown to transmit sensory signals to the bone; so called osseoperception. Osseointegration to attach the exoprosthesis was introduced by Rickard Brånemark. Infection was the main complication in a cohort of transfemoral amputees in the United Kingdom being treated using osseointegrated amputation prostheses. Based on the osseointegration concept we have developed intraosseous transcutaneous amputation prostheses (ITAP) which attempts to overcome the problems associated with infection by integrating dermal and epidermal tissues with the implant, creating a soft tissue seal around the implant[4-9]. This article is a summary of the research and development of an implant which is able to seal the skin-implant interface.

DEER ANTLER STUDY

The problems encountered with transcutaneous devices are due to the natural processes associated with wound healing where epithelial cells try to maintain continuity with one another. Around a transcutaneous device, this results in epithelial down growth, creating a pocket, which is favorable for bacterial proliferation and tissue infection. To gain an understanding of how natural transcutaneous structures are viable without the problems of epithelial layer migration and infection that are observed around artificial implants we analysed the skin bone interface around deer antlers [7]. The skin bone interface was investigated in over 20 pairs of antlers using scanning electron microscopy (SEM) of macerated specimens, transmission electron microscopy (TEM) and hard grade resin histology. Examination demonstrated a clear difference in morphology of the bone surface between the antler and the pedicle bone below the skin surface, and SEM further confirmed these findings. The surface of the pedicle is highly porous compared with the antler and the mean pore diameter larger for the pedicle compared with the antler surface. Histology and TEM demonstrated a continuous tight interface between the soft tissues and the pedicle bone in all specimens. Numerous thick Sharpey's-like fibres were observed, orientated perpendicular to the pedicle surface and emanating from the pores and spanning the dermal soft tissue-pedicle interface. The epithelial layer interfaced with the pedicle bone without signs of downgrowth. It was concluded that the sub-epithelial dermal tissue-pedical seal is critical to the success of the infection-free transcutaneous interface around deer antler and that integration of the dermal tissue with an implant may be important in maintaining an infection free interface.

DEVELOPMENT OF A BIOMIMETIC ITAP

In order to mimic the attachment of tissues seen with the deer antlers, a porous flanged structure is incorporated into transcutaneous implant inserted across the tibia in a caprine model. The porous flange structure is used to integrate and tie in the dermal tissue preventing relative motion between the skin and the implant. The transcutaneous implant was secured into the tibial with the flange positioned below the surface epithelial layer . The dermal and epidermal seal around the biomimetic implants were compared with straight pins by measuring the amount of downgrowth, epithelial layer attachment and dermal attachment. The implants remained in situ for 4 weeks after which they we removed en bloc and processed for hard grade resin histology. Longitudinal sections were taken along the length of each implant and used to quantify epithelial downgrowth, epithelial and dermal attachment. Compared to the implants without a porous flange, the biomimetic implants significantly reduced the degree of downgrowth and increased dermal attachment. Histological analysis demonstrated that complete dermal integration around the porous flange supported the overlying

epithelium, enhancing epithelial attachment and preventing down-growth.

CLINICAL TRANSLATION

The culmination of this research has led to the development of a clinical ITAP. This implant is fixed into the intramedullary cavity of the remaining bone and a porous flange under the skin surface is used to enhance soft tissue integration. This was first used successfully, in veterinary clinical cases where the animal received an amputation. Retrieval of the devices after the animal has died demonstrated good dermal integration into the flange with minimal epithelial down growth. ITAP has been used for humans with major limb loss. The first human case was of a woman who suffered multiple traumas in the London train bombing of 7th July 2005 [9]. This woman was a trans-humeral amputee and was unable and unwilling to wear a conventional exoprosthesis attached to the body using a socket and strap. She received an ITAP device that consisted of an intramedullary cementless stem partially coated with hydroxyapatite and press fitted into the diaphysis of the humerus. The rotational forces were resisted by six cutting flutes, orientated longitudinally along the distal half of the stem and which cut into the diaphyseal cortical bone. The porous flange was positioned outside the bone and below the dermis. The skin overlying the flange was attached to the implant. The muscles were sutured into a titanium mesh that was secured to the bone using cerclarge wire just proximal to the transaction site. In this way, a myodesis with the bone is achieved. After 2 years, the woman is able to go swimming and the soft tissue seal at the skin interface is entire and remains infection free. Due to the lack of a socket and straps, the range of motion that this patient achieves with her exoprosthesis is much more extensive. A clinical trial on 18 transfemoral amputees is currently under way.

FURTHER IN VIVO RESEARCH

Continued research aims to augment the epithelium and dermal seal by enhancing the attachment at a cellular level. This work has concentrated on specific surface topographies [5] and chemically coupled adhesion protein coatings [6]. We have assessed keratinocyte attachment to titanium alloy with different surface topographies using immunolocalisation of adhesion complex components including vinculin in focal adhesions, and plectin/BP180 in hemidesmosomes. TEM has been used to visualize attachment by hemidesmosomes. Smooth polished surfaces, acid etched surfaces, machined surfaces and grit blasted surfaces have been investigated. Smooth surfaces optimize cell adhesion in vitro with both hemidesmosomes and focal adhesions being up regulated compared to the rougher surfaces.

Fibronectin enhances fibroblast attachment in vitro and can be covalently attached to a titanium implant surface by silanization. The durability of attachment of this protein on a titanium surface has been measured and compared with adsorbed fibronectin when surfaces were incubated with serum. Silanized titanium alloy bound over twice the amount of fibronectin compared to untreated titanium alloy. On soaking in fetal calf serum there was no significant loss of fibronectin from the silanized surface but a significant loss from untreated surfaces. The biological activity of fibronectin bound to silanized titanium alloy was confirmed by analyzing cell area, morphology, immunolocalization of focal contacts, and metabolism of dermal fibroblasts. Fibroblasts on silanized fibronectin had significantly larger cell areas and more vinculin focal contact markers when compared to untreated surfaces. Silanization provides a durable fibronectin coating that up-regulates attachment complex expression in fibroblasts over 96 hours. These results confirm the durability of silanized fibronectin from protein competition and bioactive effect on fibroblasts [6]. A flow apparatus to assess the biophysical strength of cell attachment to biomaterials used in ITAP has also been developed. We have demonstrated that dermal fibroblast attachment strength increases significantly up to 96 h and that data from direct and indirect methods of assessing cell attachment strength have a significant positive correlation. Additionally, we have used direct and indirect assessment methods to demonstrate that dermal fibroblast attachment strength is significantly greater on fibronectin-coated titanium alloy compared with uncoated controls at 1, 4, and 24 hours.

CONCLUSION

The osseointegration concept developed by Brånemark and utilised in dental and orthopaedic applications has been developed to treat amputees so that exoprostheses can be anchored into the skeleton avoiding problems associated with fitting and transmitting loads through sockets onto soft tissues. The key issues are the fixation of the implant to the bone and importantly the creation of a soft tissue seal around the implant to prevent infection. Selecting an appropriate porous structure so that the soft tissues attach to the implant surface is important in maintaining a biological seal. In future, techniques to further enhance the formation of the seal and improve the strength of adhesion of the soft tissues with the implant may be utilised

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CASE STUDY: SURGICAL, PROSTHETIC, AND THERAPEUTIC CONSIDERATIONS FOR A PATIENT WITH IPSILATERAL BRACHIAL PLEXUS INJURY AND TRANSRADIAL AMPUTATION

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INTRODUCTION

Brachial plexus injured patients are difficult at best to treat from an orthotic or prosthetic perspective. Often times these patients present with multiple problems resulting from a flail arm presentation which may include distal "hanging" weight, lack of supporting musculature, chronic subluxation of the glenohumeral joint, scapular instability, and chronic pain [1]. Advancements in surgical management of brachial plexus injuries have resulted in greater return of functional capacity in the affected arm. Timing of reconstruction is critical as delays in surgical intervention can preclude options for successful direct repair or neurotization [2]. Delayed or late presentations, typically 3-12 months after the initial injury, can result in the need for free functioning muscle transfers for reliable elbow flexion [2]. A free functioning gracilis muscle transfer with corresponding anterior division of the obturator nerve has been described as a commonly used muscle transfer in brachial plexus reconstruction due to its proximally based muscle neurovascular pedicle and its long length [2]. The reestablishment of elbow flexion to position the hand in space should be the first priority to any reconstruction surgery. The second most important priority is stabilization of the shoulder complex [4].

Numerous orthotic and prosthetic designs have been created to accommodate and support the flaccid, paralyzed arm in an effort to provide protection and positioning to the shoulder complex, arm and hand [3]. An alternative to long term orthotic intervention that has been described in the literature involves amputating the patient's paralyzed arm at the level of an elbow disarticulation or the mid humerus in combination with a shoulder fusion to increase overall stability of the shoulder complex [1,2]. This surgical procedure usually coincides with some form of prosthetic intervention.

Shin describes transradial amputation and prosthetic intervention as a possibility when shoulder stability is maintained and when there is some elbow function spared2. He reports that even if the elbow is flail, proprioception may still be intact warranting the possibility of a more distal level amputation and some form of elbow orthosis or prosthesis that will allow for prepositioning of a terminal device in space [2].

A recent patient case within our clinical setting brought to light the consideration for treatment of this debilitating presentation. The following question was raised within our rehabilitation team: could a transradial amputation coupled with the utilization of a free-functioning gracilis muscle transfer be used to return active elbow flexion and allow for positioning of a prosthetic hand in space? Could this provide an alternative option for this patient who was considering amputation above the elbow?

CASE PRESENTATION

Patient Presentation

The patient, a 21-year old female, was seen in our clinic on August 9th, 2009, to evaluate her potential for prosthetic rehabilitation. The patient presented with a short, right transradial amputation secondary to a motor vehicle accident in Pakistan which occurred on August 8th, 2004 (Fig. 1). In addition to the right transradial amputation, the patient also sustained a right brachial plexus injury resulting in flaccid paralysis of her shoulder and arm distal to the shoulder complex (Fig. 1). A humeral fracture was also treated at this time utilizing external fixation techniques.

Upon evaluation of the patient's current physical condition, it was noted at the time that her overall health was good. Analysis of her residual limb showed good skin coverage distally but significant scarring was noted in the region of the biceps muscle. The patient presented with flaccid paralysis of her elbow with no obvious motion in flexion or extension. Her shoulder showed significant signs of wasting and atrophy and her scapula was unstable resulting in severe instability and scapular winging.

At the time of evaluation, the patient had not utilized a prosthesis or orthosis for management of her brachial plexus injury or amputation presentation. The patient's rehabilitation goals included regaining functional independence in bimanual tasks, performance of vocational activities in an office setting, and minimizing the reliance on the sound

side left hand to reduce the potential for repetitive stress and overuse injuries. Also, protection of her right arm and support of her weakened shoulder complex were very important to her continued rehabilitation.

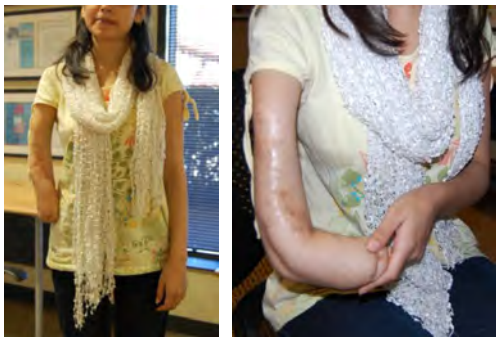

Figure 1: Initial evaluation presentation. Demonstration of the flaccid paralysis presentation

Prosthetic Recommendations

Based on her presentation, her rehabilitation goals, and her prosthetic requirements, we recommended the following prosthetic rehabilitation:

- Externally powered prosthesis
- 7 ¼" Sensor Hand Speed with stain resistant gloves
- Electric wrist rotator
- Linear transducer control of her terminal device
- Outside locking hinge with triceps cuff
- Internal lithium ion battery system
- Inner flexible socket
- Outer laminated frame with integrated locking joint
- Figure-of-8 style harness

Therapy Recommendations

The patient also required extensive therapy intervention to help strengthen her scapula and shoulder complex as much as possible. The following recommendations were made for therapeutic rehabilitation:

- Training for overall physical performance of functional movement
- Adaptive techniques training for functional activities
- Physical training for optimal UL stabilization, mobility, strength and endurance
- Patient family education for activity performance, adaptive ADL techniques, wound care and home exercise performance training

Surgical Recommendations

Because of the patient's transradial amputation presentation, we collaborated with a local plastic surgeon that specializes in brachial plexus injuries to see what options the patient might have to regain some functional elbow flexion. Based on the late presentation and severity of her brachial plexus injury, there was no potential to perform a primary nerve repair or interposition nerve cable grafting. The surgical recommendation was as follows:

- Perform a staged surgery that would ultimately use a free-functioning gracilis muscle transfer to provide active elbow function.
- The main goal of this surgery was to provide the patient the ability to pre-position a prosthetic hand in space to increase her overall function and ability to perform activities of daily living.

SURGICAL PROCEDURE

The surgery required a two stage approach. The first stage was performed in February of 2010 in which a sural nerve graft was attached to the accessory nerve in the patient's right neck. On August 31st, 2010, the patient underwent a second surgical procedure to her right arm with the goal of providing active elbow function. The procedures performed in the second stage surgery were the following:

1. Harvest of the right myocutaneous gracilis free flap.
2. Exploration of the right axilla and upper extremity in preparation for the functional muscle transfer.
3. Tenotomy and tendon repair of the right pectoralis major muscle.
4. Tendon repair of the free gracilis to right coracoid process with large Mitek suture suspension.
5. Tendon repair of the distal free gracilis to distal biceps tendon with a Pulvertaft weave.
6. External neurolysis of a previously placed sural nerve graft and neurorrhaphy to the free gracilis nerve branch.
7. Microscopic anastomosis of the posterior circumflex humeral artery to the free gracilis.
8. Microanastomisis of the brachial vein to the free gracilis.

RESULTS

Surgical Results

In the nine months following her surgery, the patient has experienced a noticeable improvement in her ability to

flex her elbow. Upon measurement of active elbow flexion, the patient has approximately 30 degrees of elbow flexion against gravity with compensatory internal rotation. The patient does not exhibit active external rotation at this time. It is predicted that the patient will gain even more elbow flexion as her free functioning muscle transfer continues to heal and re-innervate.

Prosthetic Results

Since the patient's surgery in August of 2010, we have been working on a prosthetic design that would provide functional support of the patient's arm and forearm without causing subluxation of her glenohumeral joint or exacerbate her weakened shoulder condition. During the fitting process, we were able to observe the increased function of the elbow secondary to the free functioning muscle transfer.

Our initial preparatory prosthesis was modified to accommodate a post-operative splint and sling (Fig. 2). This device utilized an Otto Bock Sensor Hand Speed with Program number three for single input proportional open and close using a linear transducer.

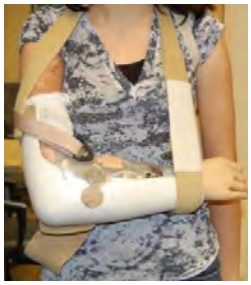

Figure 2: Initial preparatory prosthesis with post operative arm sling to immobilize the elbow in 90 degrees of flexion

As the patient healed and was able to remove the 90 degree elbow splint and sling, we moved to a second preparatory prosthesis (Fig. 3). We attempted to fit this device using short triceps cuff, an anterior Y-strap for suspension, and an axilla loop for control of the linear transducer. An outside locking elbow joint was used for control of the elbow. The patient was able to take this device with her for trial use and within a few hours of wear, her glenohumeral joint began to subluxate causing increased pain and discomfort.

Two main issues were noted in this preliminary design. The first was a lack of posterior humeral containment which allowed the arm from the glenohumeral joint down to migrate posteriorly (Fig. 3, left picture). This posterior arm migration was caused from anterior displacement of the humeral head which allowed for posterior positioning of the elbow. This created stress to the anterior glenohumeral joint which was secondary to the instabililty seen at the scapular and glenohumeral joint. The change in proximal socket design provided better approximation of the head of the humerus in the glenoid fossa, relieving anterior pressure on the joint, thus minimizing the effects of subluxation.

Second, the harness crossed over the anterior shoulder musculature creating pressure in an area of potential trigger release for the shoulder musculature (Fig. 3, right picture). This was confirmed by a replicable pinpoint pressure that created the same effect the harness did after one hour of wear time.

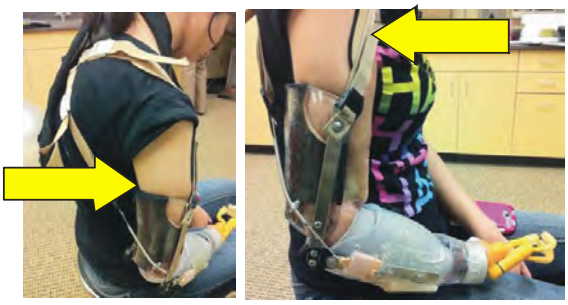

Figure 3: Second preparatory prosthesis. Notice the arrows pointing at the two main areas of concern in this design.

A third socket design was created for the patient to use that provided better overall support and stabilization of the entire arm, especially along the posterior aspect of the humerus. Also, we incorporated an anterior strap similar to that of a Wilmer Carrying Orthosis™ (WCO) to help distribute the weight of the distal forearm and terminal device5 (Fig. 4). The intent of the WCO device is to use the weight of the hand to lever the humerus into the glenohumeral joint.

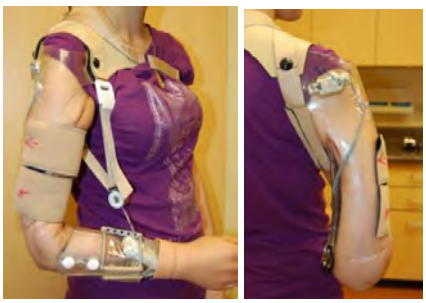

Figure 4: Better support along the posterior aspect of the humerus and overall distribution of distal weight

Ultimately, the design seen in Figure 4 worked well for the patient and we proceeded with definitive fabrication of this device (Fig. 5).

Another major change was the replacement of the linear transducer with a myoelectrode along the posterior deltoid

muscle. The control of the Sensor Hand Speed achieved by the use of the myoelectrode worked better and eliminated the need for the patient to provide excursion to the system which appeared to exacerbate her shoulder instability.

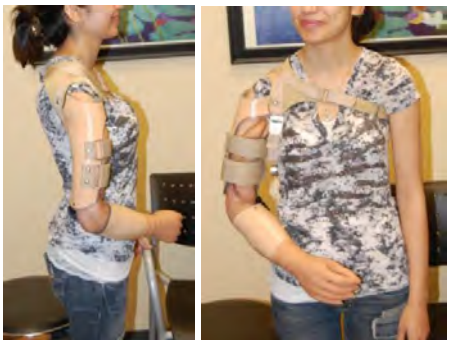

Figure 5: Finished device

After evaluation of the finished device, we considered changing the terminal device from the Sensor Hand Speed to a System 2000 hand from Otto Bock. The patient found the lighter terminal device to be more comfortable and she was able to tolerate the weight for longer periods of time. A custom made quick disconnect wrist device was created in-house for use with a MyoRotonic wrist rotator in conjunction with the System 2000 hand. This allowed for active open / close of the hand and supination / pronation using impulse control with MyoRotronic processor.

Therapy Results

Recovery from nerve graft and muscle transfer is ongoing and will continue for many months. Functional training will continue to progress with gradual improvements in muscular strength, endurance and active range of motion. Therapy intervention will continue to focus on adaptive techniques training for use with the prosthesis as an assistant for bimanual functional activities.

Interestingly, the patient began to develop referred pain from pressure points in the infraspinatus region in her right shoulder (Fig. 6). Also, the patient's scapular weakness is an ongoing problem that results in severe winging and instability in external and internal rotation (Fig. 6).

DISCUSSION

As surgical techniques improve and the ability to return active muscle function to patients suffering the effects of a brachial plexus injury, we may find that higher levels of amputation coupled with shoulder fusions are less optimal. Retention of the elbow and the ability to return function to this joint seems like a viable method of treatment. This case presentation provided to us the ability to investigate the possibility of a transradial amputation coupled with a free-functioning muscle transfer as a viable alternative to complete amputation above the elbow.

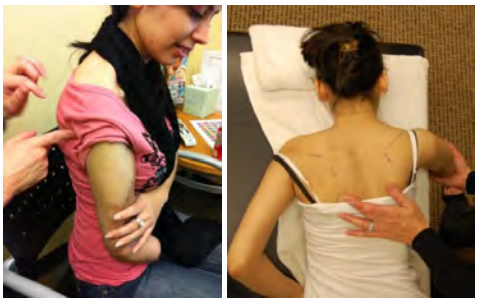

Figure 6: Demonstration of trigger points creating distal anterior pain in the arm. Severe scapular winging on the right side is noted.

As this patient develops more return in the function and strength at her elbow joint we may find that the current prosthetic socket design can be modified to a lower profile. Due to the nature of the patient's injury, concomitant muscular limitations will likely continue to limit full active range of motion at the shoulder. The prosthesis design does appear to support shoulder deficits and supplements the patients post surgical strengths for effective prosthesis use. Whether or not the patient will gain enough strength to independently lift a distal terminal device is yet to be seen.

ACKNOWLEDGEMENTS

I would like to acknowledge Dr. Jonathan Cheng for his perseverance to help our patients achieve maximum function.

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OSSEOINTEGRATION ON UPPER LIMB AMPUTEE. PROSTHETIC TREATMENT.

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BACKGROUND

Osseointegration has been used for prosthetic fixation since nineteen sixties. It is for example used in the dental and maxillofacial science [1,2]. In 1990 started the osseointegration (OI) programme for upper extremity in Sweden. The treatment involves two surgical procedures and results in bone anchorage attachment of prosthesis [1,2]. On upper limb has this method been used for transhumeral- (TH), trans-radial (TR), thumb- and partial hand amputation [2,3]. The method is also used for lower limb amputee [4-12].

Several papers and presentations in this topic have been presented at conferences and journals over the years. The aim of this presentation is to show some of the differences of an OI prosthesis compared to socket prosthesis. Procedures, constructions/fabrication, and function parameters will be lift up and how does OI effect on the Prosthetists role.

METHODS

Treatment

All patients have to pass a team assessment to find out if they are a candidate for OI. If all parts are finding OI a good solution for the patient, starts the OI treatment. The treatment involves two surgical procedures (S1,S2). At S1 is a titanium fixture inserted. Thereafter starts a healing period of normally six months [2,4,7,11]. Over this period can the patient with some limitations use the ordinary socket prosthesis [3]. But specific socket modification is most often necessary. At S2 surgery, the implanted fixture is re-exposed and the abutment is connected to the fixture. The wound is closed with the abutment penetrating the skin [2,4,7,11]. A platform for prosthetic suspension/fixation is created. The prosthetic procedure starts some weeks after S2. Initially with a lightweight prosthesis or a special training prosthesis, where the load/weight can be increased over the time [3]. This part of the treatment is depending of amputation level and type of final prosthetic. Parallel to the prosthetic treatment, implement the Occupational Therapist training and rehabilitation according to the protocol [3]. Follow-ups are carried out frequently.

Components and constructions

Together with osseointegration comes some new prosthetic components and terminology [3]. The "Attachment device" is built in to the prosthesis. Achieves a quick connector and locking function of the prosthesis and keep the prosthesis fixated to the implant. It is easy to don and doff the prosthesis. For TH and TR amputation levels is the "Puck" one part of the attachment device. The puck makes it also possible to handle individual abutment configurations on TR level, where abutment is used in both radius and ulnae. TH amputation level requires components to protect the implant from overload in rotation/torsion. For this is a "Rotation safety device" used. This component is also used for prepositioning of the forearm. Some prosthetic elbow joints on the market already include a reliable rotation/torsion function. In case of myoelectric control is "Electrode holder" used to keep the emg-electrodes in right site against the muscle position. "Alignment component" is used to optimize the prosthetic alignment. If needed can "Temperature insulator" and "Shock absorber" to avoid unwanted shock peaks or forces, be built in to the prosthesis. Some cases of TH levels need a "Soft tissue support" to stabilise the residual limbs distal tissues. A "Distal cap" can be used for protection when the prosthesis is not worn. Except from those components could selected prosthetic components on the market be used to build the prostheses. The patients can be fitted with prostheses of various types, i.e. cosmetic, body-powered, myoelectric including multifunctional and hybrids. A harness is never used for suspension but is needed for cable operated prosthesis, therefore has the TR level prostheses a built in "Wire/cable guide".

The Prosthetist role

How does this treatment affect the prosthetist role? One of the Prosthetists main goals is normally to create a good prosthetic suspension via a socket and sometimes in combination with harness. This construction shall hopefully include good function and comfort. In case of osseointegration is the suspension/fixation already ensured. The bone-anchored prosthesis always fits. It is attached correctly and is firmly held in place by the titanium implant. This eliminates all socket and harness related problems such as heat, sweating, chafing or discomfort. Change of the

residual limb volume is not an issue and the Prosthetist can spend more focus on the prosthetic function and component technology [3].

Involved in the osseintegration treatment, the Prosthetist has to be trained and learned to:

- Know how the osseointegration principal works.
- Observe the patients OI status. Take an active role in information flow in to the team and to be a part of a team.
- Supply the patient with an adequate prosthetic construction that guarantees the patients a safety situation. Never experimental construction that can risk the implant.
- Give the patient correct and relevant information regarding both prosthetic use and times when not wearing prosthesis.
- Follow the prosthetic and rehabilitation protocol, including checkups.
- In some cases, be a part of an assessment-team.
- And, listen to the patient, "listen to the bone" (PI Brånemark)

Direct bone-anchored prostheses always fit and have long durability. The need for prosthetic replacement is not frequent and worn-out sockets are no longer an issue. This reduces the prosthetic cost over a long time period. Ordinary prosthetic component services are of course not reduced.

With a fixed reference points can, alignments, prosthetic length and electrode site placement be stored. By saving those data, can prosthetic duplicate be produced and compared to socket prosthesis without a need of impressions and checkout-sockets. The patient can have a finished prosthesis delivered directly.

RESULTS/OUTCOME

Prosthetic

Different prosthetic types have been made and used in combination with OI. Attachment devices and special components have made it possible to provide patients with cosmetic, body-powered, myoelectric including multifunctional and hybrid prosthesis. Without any stump volume depending socket will the direct bone anchored prosthesis last for a very long time [3].

Patient

Approximately 40 patients have operated on upper limb. Different amputation levels have been treated, TH, TR, partial hand and thumb. OI has been successful for both short and long residual limb on TH and TR level. Causes of amputation have been trauma, congenital deformities, and tumour [3].

Function/experience

The prosthetic situation is improved because of the stable fixation. There is no need of harnessing in aim of suspension and the patients achieve full freedom of movement in the proximal joint. Problems with excessive irritation and sweating from the harness or socket do not exist. Without socket is higher degree of comfort achieved. The patients report improved functionality [3]. Clinical follow ups of prosthetic users, show improved quality of life compared to the situation before osseointegration. Patients experience improved sensory feedback because of the phenomenon of Osseoperception [13-16]. New prosthetic technology includes different platforms, where osseointegration is one important part.

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APPLICATION OF HAPTIC FEEDBACK FOR IMPROVED PROSTHETIC CONTROL

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INTRODUCTION

Tactile sensory feedback or haptics is a fundamental element of life. While feedback is vital for interaction with the outside world, current commercially available prostheses do not provide a formal mechanism to convey sensory information (1-3). The current study investigated four fundamental issues relating to external vibrotactile stimulation namely: optimal tactor location on upper arm, feedback signal type, skin desensitization, and the ability of feedback to assist in controlling grasping force. (Fig 1) A total of seven unilateral upper limb amputees participated in this study. Results demonstrated optimum feedback resolution in bicep region based on comfort and effectiveness. The average time for skin to become desensitized was 66 seconds. Among different waveforms tested, the sinusoidal waveform was the most effective (paired t-test, $p=0.047$). The cognitive loading test results demonstrated an improvement in grasping force due to haptic feedback at 60% of maximum grasping force ($p<0.05$). The preliminary haptic feedback device enhanced grasping force accuracy at specific forces rather than across all forces. (Fig 3)

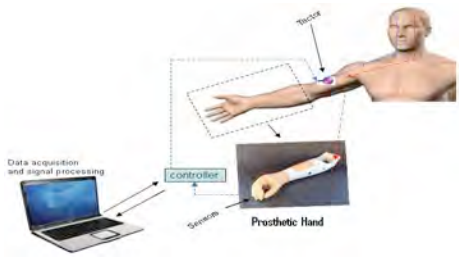

Fig 1 experimental setup for feedback testing control.

The results from Phase I of the study included clinical observations and patient feedback to provide a valuable platform toward development of a modular, customizable and clinically usable haptic feedback device in Phase II. (Fig 2)

Figure 2: some of the tests used to determine the best location and type of feedback

Different locations were tested for sensitivity to the feedback signal and then different frequencies of three different waveforms (square, sine, and sawtooth) were used to find the most effective feedback signal for each individual as they attempted to match a percentage of their maximum grip force. (Fig 3) We then tested the time for loss of sensation due to desensitization to determine how long the signal will be effective as feedback.

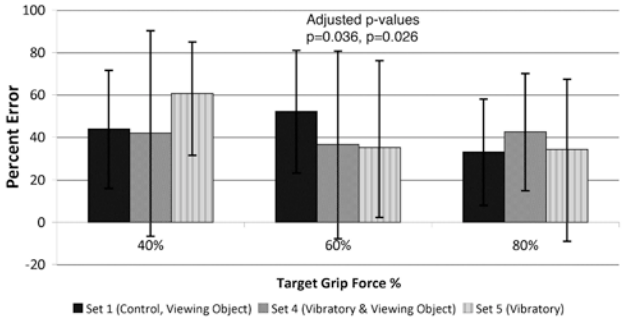

Figure 3: grip force accuracy

SUBJECT QUESTIONNAIRE RESULTS

The subjects' perception plays an important role in acceptance of assistive devices(4). As part of this study subjects were asked a set of questions relating to overall comfort, ability to use haptic feedback for daily grasping tasks, confidence in using haptic feedback and usefulness if such a device was commercially available (5-8). They were asked to rate their responses as a score from 1(worst) to

5(best). Subjects stated that feedback was helpful to improve the function of the prosthesis and that it did not decrease the comfort of the prosthesis. (Fig 4)

1. Subjects gave the highest scores to level of comfort using haptic feedback for grasping tasks followed by usefulness of such a device for everyday grasping tasks
2. They felt comfortable in using myoelectric controller and reacting to haptic feedback at the same time.

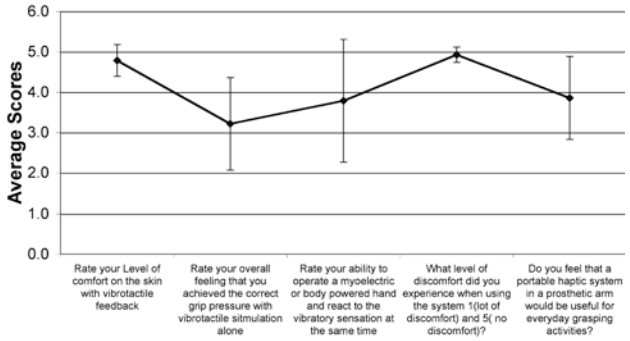

Fig 4: subject feedback on haptics outside of the lab.

RESULT ANALYSIS

1. The results showed an improvement in grasping force due to haptic feedback at 60% of maximum grasping force for Set 4 (visual and vibratory) (p=0.036) and Set 5 (vibratory only) (p=0.026).
2. Subjects who are adept at using their prosthesis (myoelectric or mechanical) were better able to utilize feedback to improve controls

The percent error while using haptic feedback improved from day 1 to day 2 at the 80% force level (p=0.007).

This indicates that more practice in using vibratory haptic feedback may further reduce gripping errors. (Fig 5)

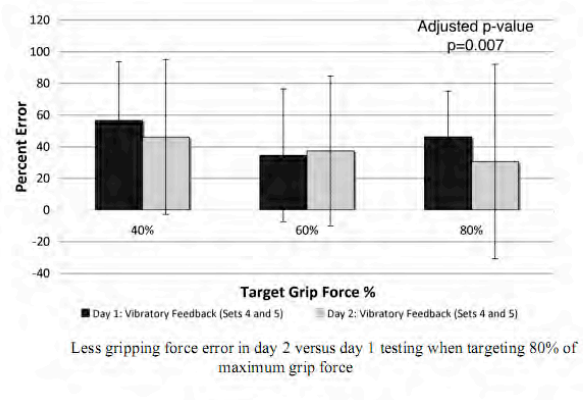

Fig 5 Improvement of grip accuracy with practice

SUBJECT FEEDBACK

1. Perceived that vibratory feedback would be helpful for activities of daily living.
2. More training with vibratory feedback would improve ability to use the feedback correctly.
3 Having three distinct force levels instead of continuous feedback would be more useful.

CONCLUSIONS

The results from Phase I including clinical observations and patient feedback have provided valuable information for development of a modular, customizable and clinically usable haptic feedback device in Phase II.

The engineering aims in Phase II consist of the design, development and integration of a low profile hardware system with wireless sensor and tactor modules. This will allow for the optimal tactor placement within the socket. Grip force and haptic feedback will be measured during common daily grasping activities to determine the effectiveness of the system and the prosthetic arm usage. Software controls will be developed for patient training and clinical use by the prosthetist to provide the most useful feedback signal. Occupational and functional measure will be used to evaluate the robustness and effectiveness of our haptic feedback system for prostheses in the lab and real world environments.

While it is virtually impossible to recreate the level of awareness of an anatomically intact limb in a prosthesis, additional sensory information through external feedback could provide a limited but valuable level of limb awareness and improved function.

The contents of this paper presentation were developed under a grant from Department of Education, NIDRR grant number H133S090044. However, those contents do not necessarily represent the policy of the Department of Education, and you should not assume endorsement by the Federal Government

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FITTING & SUSPENSION TECHNIQUES FOR A TRANSHUMERAL AMPUTEE WITH BURN INJURIES: A FOUR YEAR RETROSPECTIVE CASE STUDY

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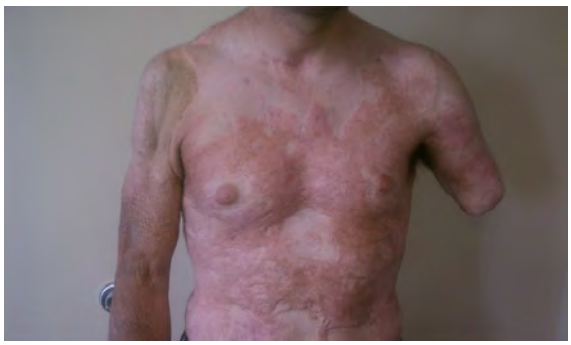
Advanced Arm Dynamics, Inc.

ABSTRACT

A firefighter with burn injuries demonstrated fitting challenges regarding comfort and suspension of a functional upper limb prosthesis. This case study introduces a transhumeral-transfemoral patient who was injured during a volunteer firefighting incident. The patient sustained several injuries which resulted in amputation of the right leg just proximal to the knee and a transhumeral amputation of the left upper limb. The patient has full range of motion in both shoulders and elbows, yet the strength and dexterity of his right hand has been compromised.

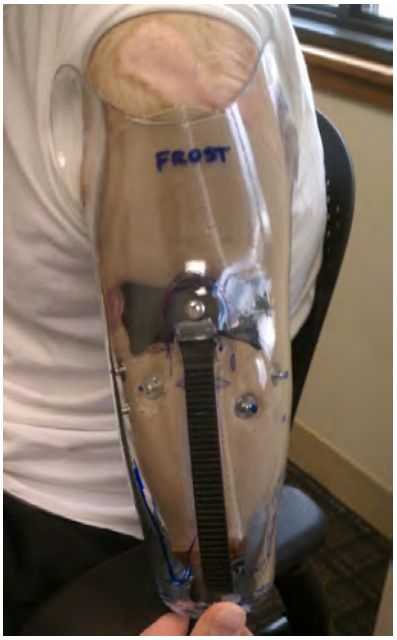


This individual sustained burns to 60% of his body, including the skin on his left transhumeral residual limb. Due to the delicate nature of his skin, a suction socket was contraindicated. Over the next four years, different types of custom and non-custom locking and cushion liners were utilized for patient comfort and suspension of both a myoelectric and a conventional (body-powered) prosthesis. The integration of electrodes for dual site myoelectric inputs offered additional challenges that compromised the suspension provided by silicone suction. Shuttle locks, lanyards, and proximal locking mechanisms were used, and the advantages and disadvantages of each system will be compared.

This case study will follow the progression of suspension techniques, interface designs, and other clinical challenges faced by the patient and the clinicians involved with the fitting. As the patient progresses toward his imminent evaluation for Targeted Muscle Reinnervation surgery, his existing design must be modified to allow additional EMG sites. The challenges of this firefighter's progressive fitting and treatment will be detailed in the discussion.

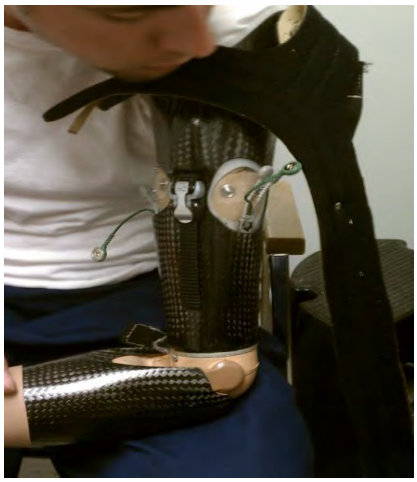
CASE STUDY

The patient was injured in August of 2007. While fighting a fire in a townhouse, the second floor collapsed and the debris trapped him. He sustained 3rd and 4th degree burns on 60% of his body and spent time hospitalized for 24 reconstructive surgeries. The patient's physiatrist prescribed a prosthesis without a harness due to the condition of the skin. The patient also was not interested in utilizing a harness due to its perceived restriction of range of motion.



The patient was initially fit with an externally powered prosthesis with a Dynamic Arm, wrist rotator, and Sensor Hand Speed. As the patient's residual limb presented with burn scars and grafted skin, initial attempts of a skin-fit suction socket were rejected. For the initial interface, custom silicone liners were fabricated for the patient. This medium provided comfort and optimal linkage. The cushion liners were fabricated with circumferential silicone rings on the

liner for suction seals. Although initially successful, frequent volume changes proved to be problematic for long term suction suspension.



The patient requested a mechanical lock be added to the liner for secondary suspension. An air tight shuttle lock was added to the distal end of the liner to supplement the suction seals on the liner. Again, volume fluctuation caused problems with the suction seals. The shuttle lock added a very subtle length discrepancy, but provided the patient with the confidence of having a secure linkage to the prosthesis.

As the patient progressed through his initial myoelectric prosthetic fitting into his definitive prosthesis, shrinkage of the residual limb required smaller liners be provided. Initially, the patient used a 20cm locking liner. He later used a 16cm liner, and eventually lost enough volume to fit comfortably into a 12cm locking liner.

One of the challenges presented while using the non-custom locking liners with holes cut at the electrode locations was the distal migration of the liners following perspiration. This migration would occur after only 20 minutes of wear time in the summer. Ultimately, the liners were changed to Alpha small uniform locking liners. To address the challenge of the cut holes in the liner, Motion Control snap electrodes were to used eliminate these openings in the liners.

As this required a new socket and frame, the shuttle lock was replaced with a medial 1" Dacron lanyard system for suspension. The lanyard suspension, also anchored from the distal end of the locking liner, did not solve the issue of the distal migration of the liner/socket.

In an attempt to reduce this tendency, a Coyote ratchet lock was added to the proximal lateral locking liner. This addition was successful in reducing the distal migration of the liner/socket. As an added benefit, the ratchet lock prevented any rotation of the residual limb within the prosthesis.

DISCUSSION

This challenging case study is important because it addresses two critical elements, suspension and myoelectric control. Compromised skin integrity as well as frequent volume fluctuations made the clinical choices less obvious.

For suspension, the initial choice of the custom silicone liners with suction seals would have provided the best linkage between the prosthesis and the residual limb, but the volume changes caused the subject to request the additional positive locking mechanism. The Ossur locking liners were more apt to migrate distally along the skin than the mineral oil-based Alpha equivalent. In both the Ossur and the Alpha locking liners, the holes that allow contact between the skin and socket-mounted electrodes allowed distal migration of the liner.

The main challenge with respect to myoelectric control was using electrodes that allowed the liner to remain in place without migrating. The snap electrodes were a viable solution for this patient; however, they must be secured to the surface of the liner or they unscrew and impair the EMG signal.

CONCLUSION

This subject presented the clinicians involved with multiple fitting experiments, including the combination of a non-suction socket with myoelectric control. For this case study, the best solution was the non-custom locking liner using snap electrodes with two locking mechanisms in addition to the harness. The redundancy of the suspensory mechanisms (two locks, harness) proved effective in avoiding distal migration of the socket during the patient's vocational setting (nursing school and EMT training).

The patient has been evaluated, approved for, and scheduled for Targeted Muscle Reinnervation surgery for the summer of 2011. A future challenge will be using 4 electrodes in the new prosthesis, as his current 2-site prosthesis uses snap electrodes. There is insufficient socket coverage to add additional snap electrodes. A custom silicone socket with an internal laminated frame/embedded electrodes is the most probable initial clinical approach.

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THE PROSTHETIC HABILITATION OF A CONGENITAL, TRANSRADIAL LIMB DEFICIENT CHILD: A CASE STUDY ANALYZING THE FUNCTIONAL EFFECTIVENESS AND THE BENEFITS OF EARLY PROSTHETIC FITTING, APPROPRIATE PROSTHETIC EQUIPMENT, AND CONSISTENT CAREGIVER FOLLOW UP

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INTRODUCTION

According to Dillingham et. al. [1] there are on average 26 children born with an upper limb deficiency per 100,000 live births each year in the United States. The Dillingham study does acknowledge that previous research from other countries has indicated an upper limb deficiency rate of 50-62.5 per 100,000 live births. Evidence and opinions are conflicted regarding the functional effectiveness and necessity of fitting young, unilateral amputee children with an upper limb prosthesis, especially in regard to the unilateral congenital below elbow deficiency (UCBED)[2-13]. The difficulty in interpreting and comparing the studies' results is that there are different definitions of successful upper limb prosthetic use.

It is this author's assertion that, although these studies have value, the entire picture of a child's needs is not addressed. Not all previous studies assure that the subjects had early and proper fitting of the prosthesis, activity-appropriate and up-to-date prosthetic devices, parental support, therapeutic training, and consistent wearing schedules. In addition, these studies don't often address or evaluate symmetry of upper body muscle development, spinal alignment and proper body mechanics while completing bimanual tasks.

BACKGROUND

This case study follows a female child with a right UCBED from birth to 7 years of age. The child was born with a "normal" presentation except for the fact that she is missing her right hand and 2/3 of her forearm. She utilizes multiple prosthetic devices.

At six months of age, the child was fitted with a passive prosthesis with a semi-flexible, passive hand attachment. This allowed her to become accustomed to wearing a prosthesis and to begin to explore right upper limb movement such as batting objects with an arm length equal to her left arm. The prosthesis was instrumental in helping to maintain sitting balance and to crawl in a typical manner with normal body mechanics. Later, a prosthetic hand in the pinch grasp shape was added. The thumb could be opened manually to allow the child or her parents to place objects in the device to encourage recognition that the prosthetic hand can hold and carry objects.

The initial plan was for a myoelectric fitting to occur at 12 months of age. This would allow the child to learn how to grasp with her right prosthetic hand at the same time that she was developing grasp patterns with her left hand. At 18 months after a struggle with the medical insurance company, she was fitted with a VASI myoelectric prosthesis with a fixed wrist. The prosthesis was activated with a "cookie crusher," single site electrode on her residual forearm extensor muscles. It performed erratically for 4 months, and it was unknown whether the problem was with the device or with the child's ability. Within 4 days of the device's problem being identified and corrected, the child began to voluntarily control the prosthesis at the age of 22 months. Through therapy and parental encouragement, the child began to use the electronically controlled prosthetic hand to reach for and to grasp objects. However, she did not have proportional control of the grasp until she began to use a dual site, proportional control program at age 3. With the new setup, she was able to demonstrate control over the speed and the grip force of the prosthetic hand during functional activities. This led to improved fine motor control. At age 6, her myoelectric prosthesis was switched to an Otto Bock System 2000 hand with a manual, rotating, friction wrist which allowed her to preposition her hand for activities and, as a result, use improved body mechanics.

Although the myoelectric prosthesis was worn full time (10-12 hours a day), it could not be used during sporting activities, sandbox play and other activities that might damage its sensitive electronic equipment and motor. At 3½ years the child was fitted with a passive sports prosthesis with a Free-Flex hand. This allowed her to begin to play sports such as soccer or t-ball. As needed, the hand was removed

and easily replaced by other terminal ends such as a fixed hook for doing the "monkey bars" or pull ups, a tumbler for gymnastics, a modified Pinch Hitter for batting, and a Slap Shot Hockey device for playing hockey. These devices have allowed participation in extracurricular activities with age-typical form.

To allow the child to play the violin at age 3, another activity specific prosthesis was created to hold the bow. The custom made device has a spring to allow "wrist" motion, which is extremely important to the mechanics of playing the violin. With this feature, she is able to maintain a relaxed shoulder on the bowing side to help prevent future shoulder injury. Recently, the violin terminal device was switched to a TRS Violin 2 bow adaptor. Violin 2 is similar to the previous device but replaces the "wrist" spring with rubber bands, thus making the wrist friction more easily adjustable for the musician.

Recently, a voluntary closing body powered prosthesis with a figure of 9 harness was provided for active grip during activities that are potentially harmful for a myoelectric prosthesis, such as a dirty or wet environment. The child has found the body powered prosthesis to be difficult to use due to the shoulder and scapular movements that are required to control the prosthesis. In order to maintain cable excursion for consistent grasp pressure while the limb is moved toward the body, abnormal shoulder and scapular positions must be used. As a result, she has not used the prosthesis unless her myoelectric prosthesis has been sent away for a glove change or repairs.

RESULTS

This child has developed in a typical manner as compared to her peers during her 7 years and has no significant medical issues. She is of average size with good posture, symmetrical upper body musculature and no noted abnormal spinal curvature. Motor coordination and development appear normal in comparison to her peers. The child appears somewhat shy in new surroundings and with new people, but once she perceives acceptance, she is at ease, friendly and participates wholly. This child has many friends and appears confident. Other children and adults seem to perceive her as a typical 7 year old child once they become accustomed to her limb difference.

Function

Wearing upper limb prosthetic devices has allowed the child to do things that she would otherwise not be able to do such as negotiate the monkey bars (with assistance), play the violin, and participate in gymnastics. She has been able to develop bimanual upper limb skills and fine motor skills with reduced compensatory movements. It is anticipated that body mechanics during functional tasks will be improved further once she receives an electric wrist rotator for her myoelectric prosthesis.

Symmetrical development of upper body musculature

By using her right prosthesis as well as her sound limb for activities and being able to perform activities with proper body mechanics, upper body musculature has developed symmetrically. In addition, there are no signs of scoliosis. Added weight may be a negative to wearing a myoelectric prosthesis, but for this child, the added weight may have contributed to the strengthening of her right shoulder, upper arm, and residual limb musculature as well as contributed to the maintenance of a straight spine.

Possible prevention of overuse syndromes

The child's development of bimanual upper limb skills with reduced compensatory movements has potentially minimized the effects of orthopedic changes and soft tissue damage that may lead to Cumulative Trauma Syndromes (CTS) in the future.

Self esteem

Measurement of self esteem is difficult because of the many variables that affect it. However, it appears that wearing a myoelectric prosthesis has had a positive effect on this child's self esteem. She likes the function and cosmesis it offers and is proud of her prosthesis. Having the opportunity to use multiple prosthetic devices which allow her to participate in age appropriate activities with her peers has also helped boost her self esteem. She knows she is different, but she feels special, instead of feeling badly about being limb deficient.

DISCUSSION

No objective outcome/standardized measures were performed on this child. Objective tests would be of interest for the sake of comparison. However, the fact that fitting this child with multiple prosthetic devices has been of benefit in terms of function, symmetrical muscular and spinal development, possible prevention of future CTS and development of positive self esteem denotes success to this particular child and the child's parents, therapists, and teachers.

The successful prosthetic outcome for this child was achieved through the following:

- Early fitting: One of the main prosthetic goals for the child was to have her fitted early with an active terminal device especially since several research studies have concluded that rejection of a prosthesis is less likely if a child is fitted before 2 years of age

[3,4,6,9]. The early fitting of a passive prosthesis at 6 months of age allowed her to become accustomed to wearing a prosthesis during most waking hours. She was able to incorporate the prosthesis into her movement strategies as she was developing the ability to reach out, bat an object, roll over, sit up and crawl. This made the transition to a myoelectric prosthesis an easy one. Fitting the child with a functional myoelectric prosthesis at 18 months allowed her to develop a pinch grasp on the right as well as to begin bimanual activities at a generally age appropriate time in her development.

- Properly fitting and up-to-date prosthetic equipment: This child was fortunate to have well fitting sockets and accessibility to prosthetic care when adjustments were required. She also was able to receive the most up-to-date prosthetic components that were available for children. This included lightweight materials and small, lightweight myoelectric batteries. She did have one experience of being fitted with a sports prosthesis that would not stay on. When she attempted to use it for anything functional, it would loosen and fall off. It was of no benefit to her. Once she was fitted with a properly fitting suspension system, she quickly incorporated the prosthesis into the desired activities.
- Therapeutic training: Early childhood special education for Occupational Therapy (OT) services began in home at 2½ years of age and progressed to OT in the preschool setting at ½ years of age. This therapy taught the child to use her prosthesis more spontaneously, to develop a consistent prosthetic finger tip grasp, to learn to use vision in place of sensory feedback, to incorporate the prosthesis into bimanual activities, to develop fine motor control and self help abilities, to develop proper body mechanics, and to develop problem solving skills. Currently she receives OT at least once during each school quarter to assess how she is progressing with fine motor tasks, typing, body mechanics, and prepositioning of her myoelectric hand. Recommendations are made to the teacher and parents so that therapy concepts are reinforced in the classroom and at home.
- Full time wearing schedule: The child's prosthetic devices, especially her myoelectric prosthesis, have been treated like a piece of clothing. The prosthesis is put on in the morning and taken off at night. Assuring consistent wearing of a prosthesis and encouraging her to use the prostheses in functional ways has been extremely valuable.
- Opportunity to try multiple devices: One upper limb prosthesis cannot replicate what a natural hand can do. Multiple devices are necessary to accomplish differing tasks. Crandall and Tomhave [12] suggest that providing children with multiple prosthetic devices appears to encourage children to wear prosthetic devices for longer periods. This child has been fortunate to have the opportunity to try different prosthetic devices and as a result has been able to participate in all age-appropriate activities like her peers. If a child is not allowed to try multiple prosthetic devices, great opportunities may be lost.

Most unilateral upper limb amputees will choose to do a one-handed task with the intact upper limb just as a person with two natural hands will prefer his dominant hand to complete a one-handed task. However, when it comes to a bimanual task, the one-handed person is at a disadvantage and will need to use compensatory movement strategies to complete the task if not wearing a prosthesis. The task will be completed, but at what cost? Compensatory movement strategies which lead to improper body mechanics have the potential to create future CTS or spinal abnormalities such as scoliosis. One of the benefits of wearing an upper limb prosthetic device is the ability to perform bilateral tasks with proper body mechanics and thus help to prevent orthopedic changes or soft tissue injury.

This author was surprised to find that little research has addressed overuse syndromes in upper extremity amputees. Jones and Davidson [14] found that 50% of upper limb amputees in their study reported that they had CTS symptoms. Extrapolating from literature on overuse syndromes in the general population [15] Gambrell suggests that overuse syndromes can occur from compensatory movements and poor body mechanics associated with unilateral upper limb deficiency. In addition, Powers, Haher, Devlin, Spencer, and Millar[16] found an increased incidence of scoliosis in people with congenital upper limb deficiencies in comparison to the general population. Asymmetrical upper body muscle development, less limb weight on the affected side, and compensatory movements may contribute to the increased prevalence of scoliosis and CTS in UCBED.

An important question is: Does wearing an upper limb prosthesis full time reduce the likelihood of CTS and scoliosis? Further research is necessary to directly correlate unilateral, upper limb amputation to CTS and to compare the incidence of injury between those groups of upper limb amputees who choose not to wear a prosthesis with those who choose to wear a prosthesis. It would also be interesting to determine which type of prosthesis results in the least overuse injuries.

It appears to this author that if the child in this case study has been successful with prosthetic devices, other UCBED children should be able to attain similar success. A child

would need to have committed parents or caregivers who will reinforce a full time wearing schedule and encourage the child to incorporate the prosthesis into activities. Therapeutic training should be provided by a therapist who has experience with upper limb prosthetic training. The child should be fitted early with a comfortable, lightweight, passive prosthesis to prepare the child for future active grasp prostheses and activity specific prostheses. Assessing which child and parents are committed to making a prosthetic fit successful is difficult, but all children should be given the opportunity.

CONCLUSION

This case study demonstrates that a child with a unilateral congenital below elbow deficiency can be successful at incorporating prostheses into her daily activities if provided with multiple, properly fitting and up-to-date prosthetic options. Fitting a child early and enforcing a consistent wearing schedule with caregiver and therapeutic follow through also contributes to functional prosthetic success. Along with functional prosthetic success comes the ability for a child to participate in age appropriate activities that may lead to positive self esteem. In addition, using an upper limb prosthetic device may help a wearer to use proper body mechanics during activities. Proper body mechanics may reduce orthopedic changes in the spine and upper body joints and reduce the potential for soft tissue overuse injuries in the future.

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OCCUPATIONAL THERAPY: TRAINING POSTURAL CONTROL FOR FUNCTIONAL UPPER LIMB PROSTHESIS USE

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INTRODUCTION

Healthcare professionals working with upper limb (UL) amputees more often than not, have the privilege of working with a generally healthy patient population. Traumatic loss of the upper limb in previously active and productive persons does not remove their intrinsic motivation for active participation in life. As they recover from the abrupt change in their functional status, this innate drive may be somewhat diminished for a time. However, with a supportive environment for recovery, it is possible to return to a healthy and productive lifestyle. It is incumbent upon rehabilitation professionals to create optimal conditions for patient success. Meeting this challenge requires the coordinated efforts of a rehabilitation team focused on the dynamic functional use of a prosthesis by the primary team member-the patient.

Partial or total loss of the UL and the associated harmful impact to motor, sensory, perceptual and biomechanical systems coalesce to influence the amputee's rehabilitation process. Of these, research suggests the potential sequelae of conditions status post UL amputation include lateral curvature of the thoracic spine. This orthopaedic abnormality in conjunction with deficits in motor systems, may have a negative influence on the dynamic function of the upper quadrant. Occupational therapists experienced in treatment of UL amputation patients utilize evidence-based methods of treatment to mitigate the physical deficits impacting functional UL prosthesis use. A review of literature to examine the effects of UL amputation to the upper quadrant and a broad view of the applicable therapeutic modalities to address resultant deficits will be presented. Specific emphasis will be given to the rehabilitation team approach to dynamic postural control for UL prosthesis use in functional activity.

THORACIC LATERAL SCOLIOSIS

A non-profit organization, The War Amps, states on its patient information page for amputee health and medical issues, "There may be a tendency, due to the weight imbalance for the amputee's spine to curve (scoliosis)." [1] This statement is supported by research specific to UL amputation. A 1996 study by Greitmann, et al, finds, "Upper limb amputations cause, in correlation to weight loss, a shift of the trunk to the side of the amputation, a scoliosis with a bowing to the side of the amputation, an elevation of the shoulder on the amputation side and a torsion of the trunk."[2] Likewise, a 1965 study of 72 Finnish soldiers with UL amputation and the related late sequelae reported, "Scoliosis of the thoracic spine must be considered a characteristic deformity in upper limb amputees, based on the investigators' findings in which 92 per cent of the above-elbow amputees and 67 per cent of the below elbow amputees presented this condition clinically." This author goes on to say, "radiologically, the frequency of thoracic scoliosis was significantly greater in upper-limb amputees than in other groups (P,0.05). In all above-elbow amputees, the thoracic curve was convex toward the side of the stump."[3]

Changes in curvature of the spine have also been noted in the pediatric limb loss population. "Scoliosis is not an uncommon finding in children with amputation, no matter what the etiology. There is need for careful examination of the entire child on both initial and follow-up visits, with further evaluations and prompt institution of appropriate treatment measures whenever indicated." [4]

THERAPEUTIC INTERVENTION CONSIDERATIONS

Smurr, et al, succinctly summarize the factors influencing the amputee's physical performance when operating a UL prosthesis. Of these, gross motor effects are highlighted. "Gross motor refers to range of motion and body symmetry. After limb loss, the client frequently compensates with shoulder elevation on the affected side." [5]

Changes in curvature of the spine and subsequent shoulder elevation stand to have a negative effect on the function of the upper quadrant and utilization of a prosthesis via musculoskeletal changes when engaging the scapula for UL activity. Additionally, as quoted in Rehabilitation of the Hand and Upper Extremity, "...early elevation of the scapula is a sign of scapular compensation for a weak rotator cuff and/or a stiff glenohumeral joint capsule. This shrugging

motion has been associated with increased upper trapezius activity."[6] The smooth activation of muscles acting on the scapula to elicit movement at the humerus is sometimes referred to as glenohumeral synergy, or scapulohumeral rhythm. A malfunction of this action can be referenced as scapular dyskinesis. "Scapulohumeral rhythm is the coordinated and synchronous movement of the shoulders osseous structures driven by the muscular and ligament systems."[6] In addition to training related to the systemic insults related to amputation, occupational therapists are trained in the assessment and effects of shoulder dyskinesia on functional use of the UL.

Research is limited regarding UL amputation's resultant affects of stresses to the thoracic spine and related treatment techniques to mitigate detrimental effects. Corio, et al, completed a study of individuals with LL loss on the effects of spinal stabilization exercise on the spatial and temporal parameters of gain. This study suggests spinal stabilization exercise training may be effective in improving selected spatial and temporal parameters of gain as a part of an overall rehabilitation program in individuals with lower limb loss through strengthening of the core muscles of the trunk, especially the transverse abdominis and multifidus."[7]

Research specific to a neurophysiological basis of trunk control in adolescent idiopathic scoliosis reveals, "Trunk control is generally carried out by means of very fast, feedforward or feedback driven patterns of muscle activation which are deeply rooted in our neural control system and very difficult to modify by training."[8] They proposed augmenting rehabilitation via bracing as a method of continuous sensory stimulation that could help awareness of body misalignment as sensory feedback.

The application of the knowledge gained from studies such as these may enhance treatment protocols to meet a patient's ability to dynamically manipulate a UL prosthesis for functional use.

THERAPEUTIC INTERVENTION OPTIONS

Research related to intervention methodologies to mitigate effects of thoracic lateral scoliosis and upper quadrant function specific to UL amputation is limited. However, therapists may apply evidence based therapeutic interventions known to be effective in treatment of the known sequelae of deficits status post UL amputation.

Thorough patient evaluation includes assessment of the spine and the dynamic function of the scapulae. Fundamental treatment methods include musculoskeletal strategies for optimal range of motion, strengthening, conditioning, neuromuscular training with repetitive drills and dynamic functional activities and psychosocial intervention and adaptive techniques training. Occupational therapists may also utilize a variety of deficit specific interventions to augment this training.

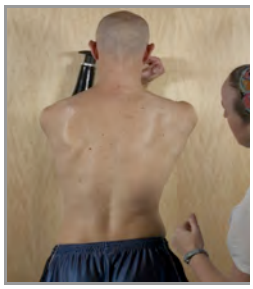

Figure 1: Spine & Scapular Stability Assessment-Trans Radial Amputee

Figure 2: Spine & Scapular Assessment-Shoulder Disarticulation Amputee

Treatment Strategies-Lateral Thoracic Scoliosis

Research related to the benefits of treatment specific to the neurophysiological effects on function of the spine support utilization of common supportive therapeutic treatment techniques with UL amputees.

Supportive treatment techniques may include virtual movement and mirror therapy to enhance cortical organization of movement. "Functional magnetic resonance imaging (fMRI) studies suggest ongoing stimulation, muscular training of the stump and visual feedback from a myoelectric prosthesis might have a beneficial effect on both cortical reorganization and phantom limb pain."[8]

Studies hypothesize the use of augmented sensory feedback and strength exercise could be an important stage in a rehabilitation program aimed at hindering, or possibly reversing, scoliosis progression.[8] Similarly, therapists may utilize kinesiology taping in addition to physical training to enhance proprioceptive and facilitory feedback. A study in the Journal of Electromyography and Kinesiology found the "application of Kinesio taping over the lower trapezius muscle improved the lower trapezius activity during 60-30° of the lowering phase of arm scaption, and increased scapular posterior tilt at 30° and 60° of arm scaption." The authors suggest Kinesio taping could be a useful therapeutic and

prophylactic assistance both in the rehabilitation clinic and in the field.[9] The concepts for utilization of this treatment technique may prove useful for UL amputee training for body awareness training.

Therapeutic intervention for pain mediation related to peripheral nerve insult is also a valuable tool for treatment of the UL amputee. Therapist attention to neural tension, neuromuscular conditioning, posture with activity and education for optimal musculoskeletal tissue healing is necessary.

UL amputees are often placed in the position of mapping new motor learning outside of their years of physical development. As such, learning new motor skills can be an exceptional challenge. Progressive repetitive training for high level dynamic prosthesis function is required for optimal motor mapping. A treatment mindset similar to that used when training athletes or musicians may be advantageous. Internationally recognized athletic training and conditioning expert, Vern Gambetta explains functional training this way, "Function employs an integrated (as opposed to isolated) approach. It involves movement of multiple body parts, and the movement involves multiple planes. It is not a matter of functional or non-functional; rather it is an understanding of how functional a particular movement or exercise is relative to the training objective." [9]

REHABILITATION COLLABORATION

As members of the rehabilitation team, it is imperative to be aware of the training and research results other team members bring to this topic. According to Donatelli, "One of the most direct relationships between the spine and the shoulder girdle is through muscle, tendon and fascial attachments."[10] Smurr, et al reports, "Coordinating therapy efforts to address the overall physical deficits associated with amputation are imperative."[5] She goes on to address the importance of partnering with physical therapists to address postural symmetry and training. "Use of a combination of methods to train for optimal dynamics performance of the upper quadrant is indispensable." [5]

The research of Yancosek, et al, highlights the opportunities to learn from the research and training of both occupational and physical therapists skilled in treating amputees. Their 2009 study on the effects of UL prosthesis use during gain in patients with concomitant LL loss highlights the effects of upper quadrant function on gait. This report surmises, "Trunk rotation and associated arm swing are critical components to human gait. Arm swing has also been purported as the motion that is useful in counteracting the trunk rotation in gait. Further, it has been suggested that abnormal trunk motion in any plane may result in decreased stabilization and poorer locomotor control."[11] This study also found that the difference between the gait pattern of the uninjured control group versus the UL/LL amputation group were fewer when the subjects wore their prostheses. This result is an example of meaningful information available for the UL amputation rehabilitation team. Patients generally perform functional tasks in a dynamic fashion such as standing, walking, bending and reaching. The use of a UL prosthesis to perform these tasks calls for an overarching team approach to rehabilitation.

CONCLUSION

It is essential to develop a comprehensive approach to rehabilitation within the context of occupational therapy and likewise the rehabilitation team. The foundation for this approach begins with the healthcare professionals working in partnership with the patient. Successful functional operation of an UL prosthesis is comprised of the coordination of intervention to address the many physical systems affected.

Further research is required to definitively ascertain the functional result of these physical insults specific to UL amputation. Longitudinal studies to ascertain the effects of UL amputation on the thoracic spine and presence of subsequent scapula dyskinesia with comparison of effects by amputation level may be of benefit. Subsequent research to identify the therapeutic treatment techniques best suited to prevent or diminish the negative impact of these effects on dynamic functional prosthesis use will enhance UL patient rehabilitation and success.

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PROSTHESIS-GUIDED TRAINING FOR PRACTICAL USE OF PATTERN RECOGNITION CONTROL OF PROSTHESES

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ABSTRACT

The potential for pattern recognition to improve powered prosthesis control has been discussed for many years. One remaining barrier to at-home use of these techniques is that practical methods of user prompting during system training are lacking. Most research and development of pattern recognition systems for prosthesis control has relied on on-screen cues to prompt the prosthesis wearer during signal collection; therefore most systems require connection to a computer or external device. We have developed a method called Prosthesis-Guided Training (PGT) to address this issue. In PGT, the prosthesis itself moves through a pre-programmed sequence of motions to prompt the wearer to elicit the appropriate muscle contractions. PGT requires no extra hardware and allows wearers to retrain, refresh, or recalibrate the controller in many locations and situations. Training via PGT is self-initiated and requires only about 1 minute of the wearer's time. Furthermore, PGT provides a practical mechanism for overcoming malfunctioning or changing inputs, addresses differences in routine donning, and results in acquisition of myoelectric signals representative of those elicited during functional use. Qualitative and quantitative data acquired to investigate the efficacy of PGT suggest that it is an intuitive, effective, and clinically viable method of training pattern recognition–controlled prostheses.

BACKGROUND

The end goal of myoelectric pattern recognition for control of upper-extremity prostheses has been successful use for a prosthesis wearer in their home and community. Previously, the use of pattern recognition for real-time control of take-home myoelectric prostheses was not possible because of the limited speed and computing power of available microcontrollers. In recent years, technological advancements have eliminated this constraint.

Many additional advancements have also been made to various elements of myoelectric signal pattern recognition systems. These include refinement of classification algorithms [1-7], improved recording electrodes [8], improvements to the stability of the electrode-skin interface [9], and development of advanced prosthetic components [10-12]. In addition, conditioning of input and output signals has been shown to significantly enhance the functionality of pattern recognition control [7, 13].

Despite the progress that has been made, there is a remaining barrier to the clinical feasibility of pattern recognition prosthesis control. This barrier stems from the fact that wearers are required to train the pattern recognition system by providing the prosthesis controller with example patterns of myoelectric signals for each desired motion. These signals are used to construct the pattern classification parameters used by the control algorithm. Frequent system training is often required, as changes in environmental temperature, limb sweating, slight limb volume fluctuations, muscle fatigue, changes to socket alignment or loading, and electrode or wire failure can all cause the performance of the system to degrade significantly, resulting in a loss of function. Without a simple and intuitive method of system retraining, pattern recognition control may not find clinical acceptance [14].

SCREEN-GUIDED TRAINING

Pattern recognition systems for prosthesis control are commonly trained using visual prompts (still pictures, text, videos, etc.) displayed on a computer screen to guide the wearer through a sequence of desired movements [2, 6, 7, 13] (Figure 1a). This is what we term *Screen-Guided Training (SGT)*. The visual cues presented during SGT provide the time sequence for recording the myoelectric signals for each motion class. Because of the long-standing popularity of SGT, it is often the only technique considered for prompting the wearer during the system training of myoelectric pattern recognition systems. Successful use of pattern recognition prosthesis control in the clinical or home setting requires an approach to system training that is intuitive and requires little to no additional hardware or technological capability.

PROSTHESIS-GUIDED TRAINING

Overview

Prosthesis-Guided Training (PGT) is an easy, intuitive method of user prompting for training and calibration of pattern recognition-controlled prostheses. The concept itself is simple: to train, refresh, or recalibrate the controller, wearers press and hold a button (a 2s hold is used in the

current system). This action prompts the prosthesis to begin moving through a short sequence of motions. The wearer watches and follows along by producing corresponding muscle contractions (Figure 1b). For example, as the prosthetic hand opens, the wearer contracts the muscle(s) that they use for the "hand open" command; when the arm stops or pauses between motions, the wearer relaxes and waits for the next movement. At the end of the sequence, all of the necessary myoelectric signals have been collected. The pattern classification parameters are quickly computed and the prosthesis is ready for immediate real-time use.

Fig. 2 Placement of the electrodes on the right hand forearm of one of the participants.

An additional processing step is required in PGT to compensate for the fact that wearers are not given advanced warning (such as the countdown typically used in SGT) prior to data collection, which produces a reaction delay. Because of this delay, it is not ideal to consider all of the myoelectric signals collected during device movement as valid training data. A mechanism has to be in place to determine when wearers begin/cease to produce the intended training signals. To accomplish this, the prosthesis remains stationary for a short period of time after PGT has been initiated (and after the prosthesis has returned to its "home" position) and myoelectric signals are collected while the wearer remains relaxed. This baseline level of myoelectric activity is used to calculate a threshold for the myoelectric signals generated during subsequent training movements (similar as in [1]). This thresholding technique has the added benefit that data collected when a wearer has forgotten to follow a motion is ignored. It also automatically provides additional training data for the "no-motion" category.

For PGT, the sequence of device movements for each wearer is the same each time they recalibrate. The wearer learns the sequence and timing of motions. This is likely to result in an improvement in the quality and repeatability of the elicited signal patterns over time, the comfort of the wearer with the system, and the number of training sessions required to produce satisfactory system performance.

Benefits of PGT

We believe there are several benefits of using PGT with a pattern recognition–controlled prosthesis:

1. *Continued wear following decreased system performance*

For many current myoelectric prosthesis wearers, the only course of action when a device stops working or begins to perform poorly is to take the device off and address the problem. Poor system performance can have a number of causes, such as broken or damaged parts, limb sweating, muscle fatigue, socket shift, and limb volume changes. Sometimes redonning the system can correct the problem; however, poor system performance often requires a visit to the prosthetist. No matter the issue, the device is generally removed or turned off, and this can occur at a time or place that is very inconvenient to the wearer. Because of this, some wearers may choose to leave a device at home. With PGT, many of these issues that arise can be overcome without having to take the arm off or even needing to know what caused the decreased system performance.

2. *No additional hardware requirement*

No external display or additional equipment is needed for PGT. When the prosthesis isn't working well (or at all), a wearer does not need to seek out a computer and display or worry about using a specific software program. Furthermore, having no computer or software means less expense to the wearer (or provider) and one less layer of technology. For system developers, graphical user interface development and software maintenance costs are greatly reduced, as is the demand for high-quality, high-bandwidth device-to-computer communication.

With the increasing popularity and presence of smart phones and personal data devices, it may be a natural fit for system developers to consider those for visual and/or audible cuing to aid in prosthetic control training and day-to-day control maintenance. Albeit more portable than a computer, using such devices still does not benefit from many of the advantages provided by PGT. Like a computer and/or display, a portable device adds cost to the system and would require the wearer to carry and maintain an additional component. Development costs with these devices are substantial, as smart phones and other devices are subject to changes outside the control of prosthesis manufacturers and developers.

3. *Fast training & recalibration*

With PGT, wearers can quickly get control of their device in the morning or after donning, and can also quickly retrain and recalibrate the device throughout the day. When a wearer dons their device after a period of non-use, they can quickly judge if they have acceptable control using what is stored in the prosthesis' microcontroller memory. If not, they may have donned their device slightly differently causing electrode shift, they may be more rested or fatigued, they may be performing contractions differently, or their skin

conditions may have changed and these changes may affect pattern recognition control of their prosthesis. In these cases and more, PGT can help the wearer recalibrate their control and resume their activities of daily living.

The prosthesis movements that wearers follow happen consecutively with small pauses between movements, meaning the whole system can be retrained or refreshed in about one minute (a 4 degree of freedom powered prosthesis). For many prosthesis wearers that system training time is potentially much less as they may only need to retrain for a limited number of powered prosthesis motions. To wearers, this means retraining can be accomplished at almost any time and place, for example, while working in their yard, in the restroom at a dinner party, during an elevator ride, or at their desk in their place of business, etc. Each wearer's strategy for using PGT can be as unique as they are, and each can find their own way to maximize their function and capabilities. All that is demanded of the wearer is that they notice a decrease in prosthetic performance and initiate the PGT refresh of the controller.

4. *Automatic normalization of dynamic range*

In most conventionally controlled prosthetic systems, careful adjustment of myoelectric signal gains, thresholds, boosts, and timings must be made by a practitioner using a computer and proprietary graphical user interface. Motion Control's ProControl II has been one of the only commercially available myoelectric prosthetic devices to have an auto-calibration feature (as described in [15]). Because PGT collects the myoelectric signals for training, settings similar to these gains, thresholds, and boosts are automatically. The collected signals are used to recalibrate the wearer's dynamic signal output range for each motion every time PGT is performed. Also, wearers often elicit muscle contractions of different intensities during PGT while following movements of different speeds. If the sequence of PGT movements incorporates a range of speeds, a larger dynamic range of myoelectric signal intensities could be acquired as training data, thereby enhancing the robustness of the control system.

5. *Increased system performance due to similarity of training and real-time use conditions*

Compared to SGT, PGT provides more similarities between training and real-time use conditions. With SGT, the wearer and prosthesis remain stationary and the wearer's attention is focused on the display and on generating distinct muscle contractions. During real-time use, both the wearer and the arm are actively moving, and the wearer is focused on the arm and the functional task at hand. The pattern of myoelectric signals produced for a distinct movement can change depending on where the arm is positioned, whether it is moving, and whether there is a load applied to the prosthesis (e.g. if the wearer is holding a heavy object or wearing heavy clothing). With PGT, myoelectric control signals are captured while the arm is moving, producing a robust classifier that performs reliably under these varied conditions. In addition, the visual and aural attention of the wearer is focused on the arm during both PGT and real-time use. This may contribute to consistency in performance between training and testing, resulting in a more functional system.

PGT in the patient education process

Although areas have been identified where the PGT method may be considered advantageous over conventional laboratory approaches (such as SGT) for training and maintaining a myoelectric pattern recognition control system, SGT approaches may remain important for initial myoelectric controls education of the patient. We believe that PGT is a clinically applicable tool for control robustness and recalibration; however, the concept of, and initial practice with, pattern recognition control will have to happen with close guidance of the therapist and/or practitioner [16]. Part of that patient education and training can be helping the wearer learn when and how to use PGT outside of the clinic.

WEARERS' FEEDBACK

Five individuals who had undergone TMR surgery [17] had the opportunity to try PGT in the laboratory setting: three subjects with a shoulder disarticulation, and two with a transhumeral amputation. All individuals used a myoelectric prosthesis and had experience with pattern recognition systems including considerable experience with SGT. Participants gave written informed consent to participate in this study.

Wearers participated in at least two separate clinical sessions where they trained their pattern recognition-controlled multifunction prosthesis using PGT. They each performed a repetitive functional task and were allowed to recalibrate their prosthesis using PGT at their convenience. In some sessions, myoelectric signal changes and disruptions were simulated in order to investigate the efficacy of recalibration by PGT. Following these sessions, wearers provided feedback via an approved questionnaire. Table 1 reports wearers' opinions on PGT. Table 2 provides some quantitative data on how wearers would be willing to retrain or recalibrate their prosthesis.

Table 1: Subjects' average responses corresponding to 5-point Likert items (1 = "Strongly Disagree", 5 = "Strongly Agree")

Questionnaire Likert-Item	Avg (Std)
I would be able to use a pattern recognition-controlled prosthesis at home if I could train it myself.	5 (0)
I would be able to notice when it is necessary to re-train ("refresh") my prosthesis.	4.8 (0.4)
PGT is intuitive; the directions are clear and easy to follow.	5 (0)
PGT is tiring.	1.8 (1.2)

Having the motions presented to me in a consistent order helps me complete PGT.	5 (0)
I would feel comfortable training my prosthesis using PGT in front of people I did not know.	4.6 (0.8)

Table 2: Subjects' average responses corresponding to fill-in-the-blank questions

Questionnaire Fill-In-The-Blank Question	Avg (Std)
I would be willing to spend up to _____ minutes to train my prosthesis each time I put it on.	5.5 (4.9)
If it were possible, I would be willing to "refresh" the control of my prosthesis while I am wearing it up to _____ times per day.	3.2 (1.7)
If it were possible, I would be willing to "refresh" the control of my prosthesis while I am wearing it no more than about every _____ hours.	2.4 (1.6)
From my experiences with it thus far, I would be willing to do PGT ____ times in a row in an attempt to get good control back instead of taking the prosthesis off.	3.2 (1.5)

The prosthesis wearers in this study became very comfortable using PGT. The wearers provided written qualitative statements on their experience with PGT:

- "When [my prosthesis] messes up, I can retrain it without taking it off. It is more convenient."
- "I learn better following the device."
- "Helps right away just by pushing a button."
- "I feel more comfortable with it […] 'monkey see, monkey do' – how easy is that!?"

An interesting observation of the wearers' experiences with PGT arose when they had to give up the PGT and return to using SGT. Most of the wearers asked for the PGT and their recalibration "button" back.

CONCLUSION

Pattern recognition control of multifunction powered prostheses may not find clinical acceptance until a very simple and intuitive method for system training is identified. We have proposed a technique where prosthesis motions are used as the cues and prompts allowing a wearer to recalibrate their control. This PGT technique may provide benefits in helping automatically adjust the control system to the wearer by overcoming day-to-day fit and signal issues. PGT also eliminates the need for additional training tools and can be accomplished by the wearer in about one minute at any time or place they are comfortable. Wearer feedback indicates very positive acceptance and desire to have PGT available.

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PROSTHESIS-GUIDED TRAINING INCREASES FUNCTIONAL WEAR TIME AND IMPROVES TOLERANCE TO MALFUNCTIONING INPUTS OF PATTERN RECOGNITION–CONTROLLED PROSTHESES

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ABSTRACT

A remaining barrier to the clinical accessibility of pattern recognition systems is the lack of practical methods to acquire the myoelectric signals required to train the system. Many current methods involve screen-guided training (SGT), where wearers connected to an external computer perform muscle contractions synchronized with a sequence of visual cues. The system complexity prevents easy retraining when signal conditions change. We have developed a method called prosthesis-guided training (PGT), where the prosthesis itself provides the cues by moving through a sequence of preprogrammed motions; screen prompting and external connections are eliminated. Five prosthesis wearers performed a repetitive clothespin placement task using pattern recognition control. Wearers demonstrated similar baseline functionality between systems trained with PGT (10 ± 4 clothespins) and SGT (12 ± 7 clothespins) ($p = 0.56$). To investigate the efficacy of PGT retraining, real-world issues (e.g. broken wires, external noise) were simulated to accelerate control degradation. Sessions ended when wearers indicated loss of functional control. On average, wearers maintained function through two malfunctioning inputs, placing 48 ± 17 clothespins in 31.6 ± 16.2 minutes when allowed to retrain using PGT. These results suggest that PGT acquires adequate training data and may enable longer-lasting functional use, potentially increasing prosthesis wear time and reducing device rejection.

INTRODUCTION

Pattern recognition–based control has shown promise for myoelectric control of upper limb prostheses but has had limited clinical implementation. For pattern recognition systems to progress and move outside the laboratory, a few remaining barriers need to be addressed. This paper focuses on developing a practical way for prosthesis wearers to train the system.

Existing methods of acquiring the myoelectric signals necessary to train the system have generally relied on visual or auditory cues. During a screen-guided training (SGT) session, wearers connect to an external computer and perform muscle contractions synchronized with a sequence of visual cues [1]. This method has seen wide-spread implementation in the laboratory but may not be practical for home use. The requirement for an external display adds to system complexity and prevents easy retraining when signal conditions change. Auditory cues can eliminate the need for an external display but rely on the wearer to remember a preprogrammed sequence of movements.

Prosthesis-guided training (PGT) is a new method that eliminates the need for an external connection. During PGT, the prosthesis provides the cues for the wearer. Wearers initiate PGT by simply pushing a button. The prosthesis then moves itself through a sequence of preprogrammed motions. Wearers follow along with the prosthesis motions by performing the necessary muscle contractions and relaxing each time the prosthesis pauses between motions. The myoelectric signals that are collected during this sequence are immediately used to train and recalibrate the pattern recognition control system.

The goal of this study was to investigate the efficacy of retraining a pattern recognition prosthesis system using PGT, Wearers can encounter several different types of issues in their home and community that can cause their prosthesis control to degrade. Faulty electrodes and changes in signal quality are two major problems that can occur during use. We simulated these real-world issues at regular intervals in the laboratory to test wearer performance during periods of accelerated control degradation. Providing wearers with an easy method of retraining and recalibrating their prosthesis if and when these issues arise can increase wear time and reduce device rejection.

METHODS

Five individuals who had undergone TMR surgery participated in this study: two male participants with a right shoulder-disarticulation (S1 and S2), one female participant with a left shoulder-disarticulation (S3), one male participant

with a right transhumeral amputation (T4), and one female participant with a left transhumeral amputation (T5). All individuals used a myoelectric prosthesis and had experience with pattern recognition systems Participants gave written informed consent to participate in this study.

Eight bipolar electrode pairs were placed on the skin surface over the reinnervated muscles. The myoelectric signals were amplified, sampled at a frequency of 1 kHz, high pass filtered (20 Hz cutoff frequency) to reduce motion artifact, and processed in real time using custom software.

The pattern recognition algorithm was trained to recognize nine motions: elbow flexion, elbow extension, forearm supination, forearm pronation, wrist flexion, wrist extension, hand open, hand close, and no movement. Six seconds of data for each motion were used to train a linear discriminate analysis (LDA) classifier [2] and six seconds of data for each motion were used to determine the classification error. The EMG data were segmented into a series of 250 ms analysis windows [3] with a 50 ms window increment.

Four time-domain values (mean absolute value, number of zero crossings, waveform length, and number of slope size changes [4]) and autoregressive coefficients were computed and used in pattern classification. After the LDA classifier was trained, it was used to predict user commands and control a prosthetic arm. The motion speed was normalized to the training data contraction intensity and a 500 ms velocity ramp was applied to minimize the effect of misclassifications [5]. This setup resulted in a clinically viable, functional system requiring no experimenter adjustments of output gains or thresholds.

Prosthesis-Guided Training vs. Screen-Guided Training

Individuals participated in two separate experimental sessions. The difference between sessions was how the myoelectric signals required to train the pattern recognition system were collected. The order of sessions was randomized. For the SGT session, wearers were connected to an external computer and performed muscle contractions synchronized with a sequence of visual cues. For the PGT session, wearers performed muscle contractions synchronized with a sequence of preprogrammed motions of their prosthesis. Wearers self-initiated PGT by pushing a button attached to their prosthesis. Both methods collected the same amount of training data.

To measure performance between the two methods of collecting training data, wearers performed a repetitive clothespin placement task. In a 4 min baseline trial they moved as many clothespins as possible from a horizontal bar to a vertical bar [6] (Figure 1). A paired t-test was used to detect significant differences in performance between the two sessions.

Figure 1: Prosthesis wearer performing the clothespin placement task.

Accelerated Life Cycle Test

To investigate the efficacy of retraining the pattern recognition system using PGT, real-world issues were simulated to accelerate control degradation. The simulated real-world issues were either a faulty electrode or a noisy electrode. The simulated faulty electrode (i.e. channel amplitude set to zero) was representative of a broken wire or faults in the electrode circuitry. The simulated noisy electrode (i.e. addition of large 60 Hz interference) was representative of external noise and/or electrode lift-off. Issues were cumulative in nature and wearers were blind to the type and timing. Wearers performed the repetitive clothespin placement task in 12 min blocks, each followed by a 4 min break. Four minutes into each block, one issue was applied to a randomly selected channel. Wearers were instructed to continuously perform the clothespin task. Sessions ended when wearers were no longer able to place clothespins, indicating a loss of functional control.

During the PGT session, wearers were able to self-initiate recalibration of their prosthesis when they believed their performance had degraded. During the separate SGT session, wearers did not have the option to recalibrate their prosthesis. This session was representative of the wearer being in an environment where the external computer necessary for SGT was not available. The simulated electrode issues occurred in the same order across sessions and the order of sessions was randomized. Performance metrics included prosthesis wear time, total number of clothespins placed, number of issues overcome, and the times between onset of an issue and initiation of retraining using PGT.

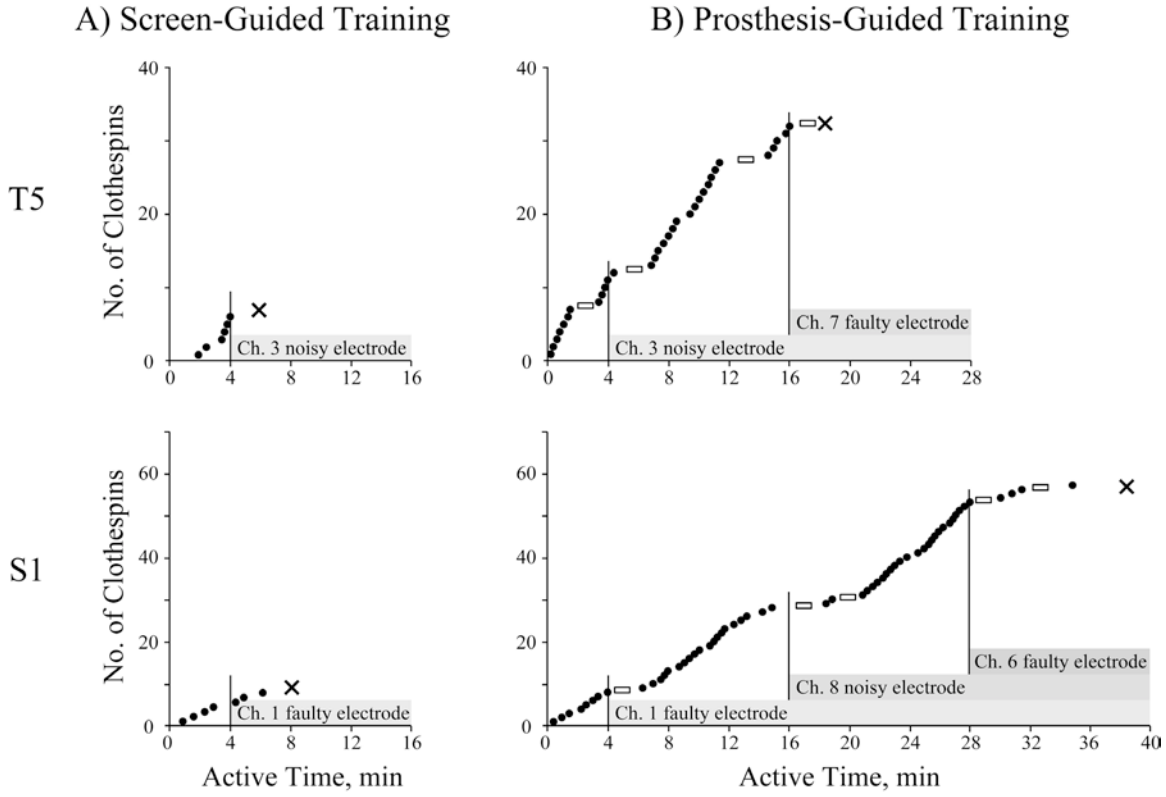

Figure 2: Number of clothespins placed vs. time for a prosthesis wearer with a shoulder disarticulation (S1) and a prosthesis wearer with a transhumeral amputation (T5) for (A) screen-guided and (B) prosthesis-guided training. The active time reported does not include the 4 min breaks that occurred every 12 min. The simulated real-world issues were cumulative in nature. Circles indicate a successfully placed clothespin, rectangles indicate self-initiated PGT (if available), and x indicates loss of functional control.

RESULTS

Prosthesis-Guided Training vs. Screen-Guided Training

Classification error for myoelectric signals collected during SGT (7.8% ± 3.5%) (mean ± standard deviation) was significantly lower than during PGT (18.1% ± 2.8%) (paired t-test, $p = 0.009$). During the baseline clothespin test, wearers demonstrated similar baseline functionality between systems trained with SGT (12 ± 7 clothespins) and PGT (10 ± 4 clothespins) ($p = 0.56$).

Accelerated Life Cycle Test

Figure 2 shows the accelerated life cycle test for two wearers. In this example, an individual with a transhumeral amputation (T5) lost functional control after one channel was affected. With SGT and no option to recalibrate her prosthesis, her only option was to take the prosthesis off (Figure 4A). With PGT available, she recalibrated her prosthesis when her control degraded. She maintained function after the same channel was affected, thereby extending the functional use time of her prosthesis (Figure 4B). An individual with a shoulder disarticulation (S1) also lost functional control a few minutes after one myoelectric channel was affected. With the option to retrain using PGT, he was able to maintain function even after three of eight myoelectric signals were affected (two channels had simulated faulty electrodes and one channel had simulated noise). This individual retrained his prosthesis using PGT five times. When pattern recognition control degraded in response to simulating either faulty electrodes or noisy electrodes, all wearers initially chose to retrain their prosthesis instead of indicating loss of functional control and taking their prosthesis off.

During the SGT session with no external computer available to recalibrate their prosthesis, wearers placed a average of 10 ± 5 clothespins in 6.2 ± 1.7 minutes. Given the option to retrain their prosthesis using PGT, wearers placed an average of 48 ± 17 clothespins in 31.6 ± 16.2 min. With PGT, wearers maintained function through an average of 2.0 ± 1.4 malfunctioning input channels. After a signal channel

was affected, wearers retrained their prosthesis using PGT within 32 ± 22 seconds. Wearers retrained their prosthesis an average of 5.8 ± 4.1 times.

One wearer (T4) was excluded from the analysis because his results were constituted as an outlier; his performance metrics for both SGT and PGT sessions were above the sum of the third quartile and 1.5 times the interquartile range. T4 did, however, show the same trend that he was able to overcome malfunctioning input channels and extend functional use when allowed to retrain using PGT. He placed 57 clothespins in 15.8 minutes during the SGT session and 104 clothespins in 42.5 minutes during the PGT session.

DISCUSSION

Results from the baseline clothespin test demonstrate similar functionality between systems trained with SGT and PGT. The myoelectric signals collected during PGT may be different than those collected during SGT. During SGT, wearers are focused on a display while performing the muscle contractions and their prosthesis remains static in the neutral position. During PGT, wearers, by design, are focused on their prosthesis. Their prosthesis is in motion, providing the cues necessary for them to initiate the corresponding muscle contractions. As the prosthesis moves, it alters socket-tissue loading and the muscle activity necessary to support the moving weight. Therefore the conditions in which these signals are collected in order to train the pattern recognition system are more similar to the environment in which wearers will use their prosthesis. These changes are recorded during PGT but not SGT. PGT may capture more transient signals as each muscle contraction is recorded from rest, which may have lead to the higher PGT error rates during offline analysis [7].

Our results suggest that with PGT, wearers may be willing and able to maintain functional use of their prosthesis longer than without it. When we simulated a broken wire or signal noise, wearers noticed their control degrade. With SGT and no external computer available, wearers lost functional control and had no other choice but to take their prosthesis off. SGT does not necessarily require an external computer and could be performed using a smartphone application. Nonetheless, equipment in addition to the prosthesis is still required and smartphones may not be available to or desirable for all wearers.

With PGT, wearers self-initiated recalibration of their prosthesis in an attempt to restore control within an average of 30 s. If they were at home and a wire broke, PGT may provide them with a longer time frame of functional use before they need to go back to the clinic. Without PGT, most likely the device would be uncontrollable. Wearers would not have the option of using their prosthesis until they could return to the clinic. For less extreme issues, such as changes in skin conditions or muscle fatigue, PGT would also offer wearers the ability to quickly recalibrate their control.

CONCLUSIONS

PGT is a straightforward way for wearers to retrain and recalibrate their prosthesis when myoelectric signal conditions change. With PGT, wearers can take an active role in trying to improve their control and attempt to overcome control issues instead of taking their prosthesis off. This study demonstrated that wearers are willing and able to retrain their prosthesis. Wearers can seamlessly transition back to the task they were performing prior to the PGT session. This new method of acquiring the myoelectric signals necessary to train a pattern recognition system has the potential to increase wearers' usage time and reduce device rejection.

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PRELIMINARY STUDY ON THE INFLUENCE OF INERTIA AND WEIGHT OF THE PROSTHESIS ON THE EMG PATTERN RECOGNITION ROBUSTNESS

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ABSTRACT

For transradial amputees, the muscles in the residual forearm naturally employed by unimpaired subjects for flexing/extending the hand fingers, are the most appropriate targets, for multi-fingered prostheses control. However, once the prosthetic socket is manufactured and fitted on the residual forearm, the recorded EMG might not be originated only by the intention of performing finger movements, but also by the muscular activity needed to sustain the prosthesis itself. In this work, we preliminary show –on healthy subjects wearing a prosthetic socket emulator– that (i) variations in the weight of the prosthesis, and (ii) upper arm movements significantly influence the robustness of a traditional classifier based on k-nn algorithm. We show in simulated conditions that traditional pattern recognition systems do not allow to separate the effects of the weight of the prosthesis because a surface recorded EMG pattern due only to the lifting or moving of the prosthesis is misclassified into a hand control movement. This suggests that a robust classifier should add to myoelectric signals, inertial transducers like multi-axes position, acceleration sensors or sensors able to monitor the interaction forces between the socket and the end-effector.

INTRODUCTION

To myo-electrically control a multi-fingered dexterous prosthesis –like e.g. the recently marketed RSLSteeper BeBionic [1] or research prototypes like SmartHand [2] or the Vanderbilt University Hand [3], it is necessary to map electromyographic (EMG) signals corresponding to different muscle contractions to the different existing degrees of freedom (DoF) of the hand using a suitable algorithm. In research this is frequently done through pattern recognition based techniques [4]. Since the 1960s, various groups have designed controllers using different combinations of extracted features and classification methods (for a review of the EMG processing techniques refer to [5]) showing the feasibility of controlling dexterous prostheses. These systems have been demonstrated usually through offline pattern recognition [6]-[8], through algorithms suitable for real-time processing and classification [9]-[11], but only in few instances, with actual real-time classifiers [12]-[14] or directly controlling robotic hand finger movements [15], [15]. Results in this field are improving incrementally but slowly, and research is mainly focusing on real-time signal processing techniques, pattern recognition algorithms and other computing issues.

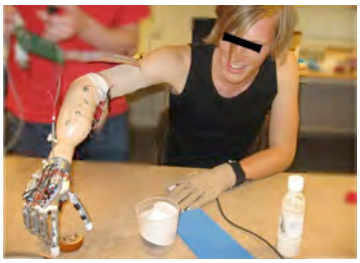

Fig. 1 Amputee reaching an object wearing SmartHand. The unnatural reaching posture of the arm caused by the lack of the 3 degrees of freedom of the wrist/forearm is clear from this picture.

However, all previous research is related to experiments performed in controlled laboratory environment, with the stump of the subjects lying in a **comfortable position**: i.e. with no moving limbs/stumps. It is foreseen that future systems should be able to deal with bio-signals coming from a **free-to-move** residual limb; in such case, the main open problems are: source localization (muscle motion problems), skin impedance changes, removal of artefacts, prosthesis donning/doffing, and separation of intention from other physical factors (like fatigue, stump posture, etc.). In transradial amputees, the (up to) 19 extrinsic muscles in the residual forearm which naturally are employed by unimpaired subjects for flexing/extending the hand fingers, are the most appropriate targets, for multi-fingered prostheses control. However, once the prosthetic socket is manufactured and fitted on the residual forearm (cf. Fig. 1), the recorded EMG might not be originated only by the intention of performing finger movements, but also by the muscular activity needed to sustain the prosthesis itself. Indeed, in contrast to an healthy forearm, for amputees, the actions caused by the weight of the prosthesis (payload and inertia while moving) are partially distributed on the muscles above the elbow (e.g. biceps-triceps), and partially on the forearm muscles; this being reinforced by the reaching posture of the prosthetized limb which is generally unnatural due to the lack of biomechanically correct wrist movements

(cf. Fig. 1). Additionally, movements of the socket relative to the stump (caused e.g. by the inertia of the prosthesis when it is moved) might generate artefacts, i.e. involuntary signal variations. Traditional techniques do not allow to separate such effects, therefore, an EMG pattern due only to the lifting or maintaining of the prosthesis can be misclassified into a hand control movement, as a consequence of a false positive.

To tackle this problem, the idea of a robust interface including EMG and inertial transducers (i.e. multi-axes position and acceleration sensors) for intuitive prostheses control was recently patented by Cipriani *et al.*, [17] and similarly, the adverse effects of limb position on pattern recognition control were investigated on healthy subjects and presented by Scheme *et al.*, [18]. Within this framework, in the present paper, we preliminarily show –on three healthy subjects and emulated conditions– that (i) variations in the weight of the prosthesis, and (ii) upper arm movements weaken the robustness of pattern recognition. Results of this work, although still preliminary, suggest a simple but effective strategy for the control of multi-fingered prostheses based on the monitoring of the prosthesis weight and upper limb posture.

MATERIALS AND METHODS

Three able-bodied subjects (two men and a woman aged 25, 27 and 27 years old, respectively) took part in this preliminary study. The dominant hand was the right hand for the first and third subject and the left one for the second subject. Raw surface EMG data were collected employing the Noraxon TeleMyo 2400R (Noraxon, Scottsdale, AZ, USA) through a wireless unit (TeleMyo 2400T). Raw data were then acquired at a sampling frequency of 1.5 kHz, 1st order 10 Hz hardware high-pass filtered, 8th order 500 Hz hardware Butterworth low-pass antialiases filters, resolution of 12 bits, hardware gains of 1000, and stored for an offline analysis in MatLab environment. In order to investigate on individual finger classification eight channels were used to record myoelectric activity from the right-hand forearm muscles. Disposable Ag–AgCl surface electrodes in bipolar configuration with an inter-electrode distance of 20 mm were used. Four channels recorded signals from superficial flexor muscles on the volar side of the forearm and four channels were placed on the superficial extensor muscles on the dorsal side of the forearm as shown in Fig. 2. The reference electrode was placed on the proximal part of the lateral epicondyle.

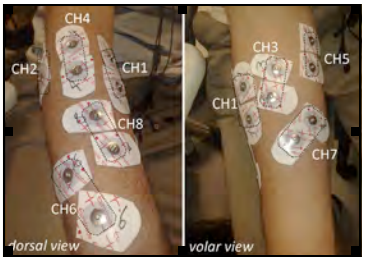

Fig. 2 Placement of the electrodes on the right hand forearm of one of the participants.

The participants were seated in front of a screen with their forearm resting on a pillow during the time of this experiment. The hand default posture allowed the extrinsic muscles to be totally relaxed, as visually inspected through the EMG recording system. Ten different movements were executed by the subjects in response to a written and pictorial cue on the screen and an auditory cue that depicted the movement to be reproduced. The movements consisted of flexions and extensions of the thumb and index fingers individually, of the middle, ring, and little finger as a group, of the long fingers (all but the thumb) as a group and of thumb abduction, and finally of a rest class making up ten classes in total. These movements would account for individual control of each degree of freedom of an advanced prototype like the VU- or the Smart- hand [2], [3]. Each movement was sustained for 5 seconds and a 5 second rest was given between subsequent movements. Two different datasets each consisting of 3 repetitions of each movement totalling 27 movements and the rest states were stored on a computer along with the intended class information.

A simple but effective classifier already used in our previous work was employed [16]. It consisted of a k-nearest neighbour (with k equal to 8) algorithm employing the Euclidean distance as the distance metric and the mean absolute value (MAV) as feature set. For both subjects the first recorded dataset was used for training (hereafter calibration dataset) and the second for evaluation. The resulting classification accuracies are shown in the confusion matrices in Fig. 3 It is worth underlining that the classification accuracy for the relax state was 91%, 95% and 89% for the first, second and third subject, respectively. Two experiments –as detailed in the following sub-sections- were carried out in order to assess the worsening effects of the weight (i.e. payload and inertia while moving) of the hand prosthesis on a simple pattern recognition based control.

Weight Effects

In order to resemble the fact that transradial amputees wear a prosthetic socket usually rigidly connected to the elbow and hence cannot pronate/supinate the forearm, subjects during this experiment wore a prosthetic socket

emulator (cf.Fig. 4A-D), that impeded forearm movements and kept the hand always in fixed –and relaxed– position.

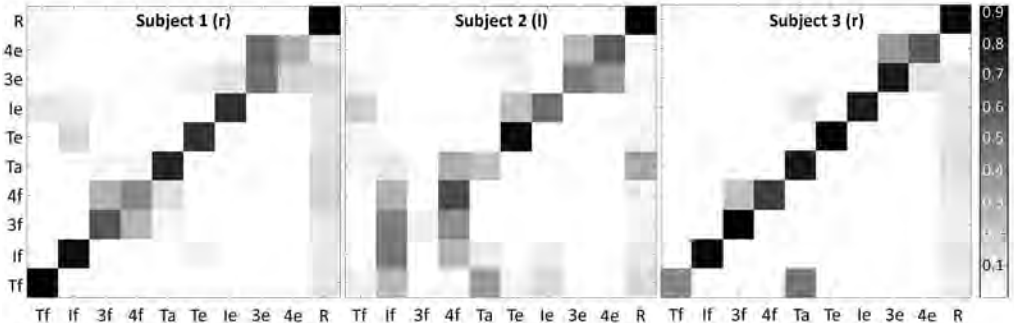

Fig. 3 Confusion matrices from the three participants. Movement list: Tf: thumb flexion, If: index flexion, 3f: three fingers (middle, ring and little) flexion, 4f: four fingers (index, middle, ring and little) flexion, Te: thumb extension, Ie: index extension, 3e: three fingers extension, 4e: four fingers extension, R: relax. The letter in brackets refer to the dominant hand.

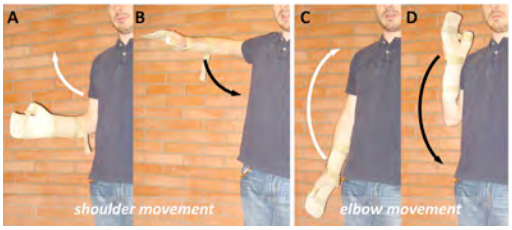

Fig. 4 Experimental protocols. Shoulder abduction/adduction movement (A-B) and the elbow flexion/extension (C-D). The postures depicted in pictures A and B were also used in the weight effects experimental protocol.

Subjects were asked to maintain a static posture with their right arm, while the endpoint of the socket emulator was cyclically loaded and unloaded with a mass (3 seconds loaded and 3 seconds unloaded, 5 times). Two static postures were tested, the first (posture A) with the arm attached to the body and the elbow forming a 90 degrees angle (cf. Fig. 4A) and the second posture (posture B) maintaining the elbow flexion and abducing the shoulder until bringing the arm in line with it (cf. Fig. 4B). Theoretically in both postures the payload was not supported by forearm muscles (those involved in the grasp action), but by arm and shoulder muscles. Subjects were instructed to keep their forearm muscles always relaxed during the loading/unloading cycles. In the first posture 3 loads (10, 15 and 20 N) were tested; in the second posture just the 20 N load was used. This protocol aimed to imitate and investigate the effects on pattern recognition of the weight of the prosthesis acting with a certain lever arm on the prosthetized stump of a transradial amputee. The recorded EMGs were classified using as training data the calibration dataset.

Movement Effects

Effects of inertia on the classification accuracy were tested in this second experiment. Subjects were asked to execute two kinds of movement not involving the forearm muscles: the first one was shoulder abduction/adduction (between postures A and B inFig. 4A-B), the second one was elbow flexion/extension (between postures C and D in Fig. 4C-D). In both cases subjects were asked to perform cyclically at physiological speed (i) the first part of the movement (e.g. shoulder abduction), (ii) keep the position for 3 seconds, (iii) perform the second part of the movement (e.g. shoulder adduction) and (iv) keep this position for 3 seconds. Audio cues for an easier synchronization were delivered through earphones. In order to mimic the prosthetized condition a 0.5 kg mass was attached to the end of the socket emulator (the standard weight of an adult size prosthesis is around 0.5 kg indeed [1]-[2]). Subjects were instructed to keep their forearm muscles always relaxed, and the EMG signals while performing the movements were acquired and off-line classified using as training data the calibration dataset.

RESULTS AND DISCUSSION

Weight Effects

Subjects were instructed to keep their hand relaxed during the loading/unloading cycles. Since the mass was ideally sustained by biceps and shoulder muscles (in posture A and B, respectively), the extrinsic muscles of the hand in the forearm were not supposed to be active. Instead, as hypothesized in the introduction the load was partially sustained also by the forearm muscles, which activity led to misclassification of the relax state. This effect is depicted in the temporal graph in Fig. 5 where a representative sample from subject 2 is shown (load: 15 N). The black line denotes the mean MAV among the 8 EMG channels, whereas the red

dots indicate the output class label computed by the k-nn classifier (label 5 corresponds to the relax class). U and L intervals on the time scale denote the load and unload phases, respectively.

The graph clearly shows the myoelectric activity variations causing the relax state to be misclassified every time the load was applied, and properly classified once the load was removed. Table 1 resumes the relax classification accuracies during the loading phases (grey windows in Fig. 5) included in the whole dataset, for the three subjects in both postures tested (cf.Fig. 4A and B). The effects of the weight were highly subjective and further investigations are hence required before being able to draft any conclusion. However, as a general preliminary remark, static loads yielded to a decreased classification accuracy (worse for subject 2 where EMGs were recorded from his non-dominant arm). By transferring this to the transradial amputee situation, a traditional pattern recognition algorithm would generate involuntary control commands every time the weight of the prosthesis changes (e.g. every time a new object is grasped).

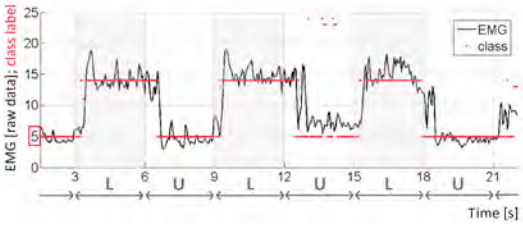

Fig. 5 EMG activity (black line) and classifier output (red dots) from Subject 2 during loading (L) and unloading (U) phases using the 15 N load.

Table 1: Classification accuracies of the relax state at different loads and limb postures

	Posture A		**Posture B**	
	10 N load	***15 N load***	***20 N load***	***20 N load***
Subject 1	100%	89%	20%	12%
Subject 2	1%	6%	1%	8%
Subject 3	100%	98%	44%	4%

Movement effects

A representative temporal graph of EMG activity and classifier output stream is shown in Fig. 6. Similarly to the other test, the plot shows that the myoelectric activity causes the relax state to be misclassified every time the forearm moves (from C to D, cf. Fig. 4C-D), and is maintained flexed (posture D). In this case the activity might also be caused by artefacts due to cyclical peaks of pressure of the socket emulator on specific electrodes; this effect would still be present in the case of an amputee wearing a prosthetic socket, hence is of interest of this study.

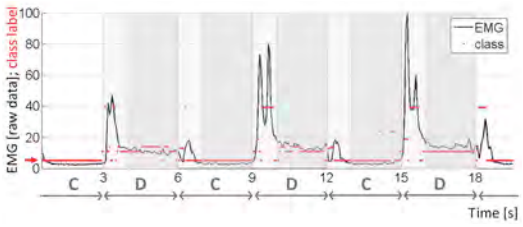

Fig. 6 EMG activity (black line) and classifier output (red dots) during flexion-extension of the elbow by Subject 2. C and D time intervals represent the windows when the elbow was flexed and extended, respectively (as in Fig. 5C and D).

Table 2 quantifies the relax classification errors resulting from the whole dataset for the three subjects performing the two movements, during the first second after the movement cue (light-grey windows in Fig. 6), and during the two subsequent seconds (dark-grey windows in Fig. 6). The former relates to the dynamic part of the movement, whereas the latter refers to the static phase.

The classification errors are considerably high, and as presumed, greater in the dynamic part of the movement than in the static one. While the reason for the misclassification in the dynamic phase can be attributed to the effects of inertia on the classifier and on the muscle-electrode interface (skin movement artefacts), the misclassification in the static phase is probably due the 0,5 kg mass attached to the socket emulator. By transferring this to the prosthetized situation, a traditional pattern recognition algorithm would generate involuntary control commands every time the prosthesis is moved.

Table 2: Classification errors of the relax state with different movements

	Shoulder movement		**Elbow movement**	
	Dynamic	***Static***	***Dynamic***	***Static***
Subject 1	39%	38%	43%	4%
Subject 2	15%	12%	45%	19%
Subject 3	40%	25%	24%	17%

To obviate this clinical issue once the socket is fitted on the stump, i.e. to remove the load and inertial effects of the prosthesis on the amputee's residual forearm, one possible approach is to monitor the posture and movement of the prosthetized limb (this data could be easily computed by means of DoF sensors, having on board accelerometers and gyros along multiple axis) and/or monitor the interaction forces between the socket and the prosthesis (by means of multiple axis load cells). Such information could be used to compute the load and inertial force vectors which affect EMGs, and once modelled, such effects could be compensated by the controller.

ACKNOWLEDGEMENTS

This work was supported partially by the Ministero dell'Istruzione, dell'Università e della Ricerca, under the PRIN 2008 OPENHAND Project prot. PMZT2Z and partially by the Cassa di Risparmio di Lucca Pisa e Livorno under the research project on upper limb prostheses.

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FEEDBACK IN VOLUNTARY CLOSING ARM PROSTHESES

Investigation of optimal force feedback in shoulder controlled arm prosthesis operation

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INTRODUCTION

Commercially available arm prostheses do not fulfil the requirements of the users. Twenty to forty percent of arm amputees choose not to wear a prosthesis. Of those who wear a prosthesis, roughly half does not use the full functionality it offers. Instead, they use the prosthesis for its cosmetic function [1].

A prosthesis should look natural, be comfortable to wear, and easy to use [2]. Unfortunately, shoulder controlled body powered prostheses require high operation forces, resulting in discomfort and fatigue of the users [1, 3], which results in high rejection rates [4].

The ease of prosthesis control depends (among other things) on the necessity of watching the operation of the prosthetic prehensor, to prevent slipping or crushing of the object grasped. Eliminating the need for visual monitoring the operation will lead to subconscious control, therewith decreasing the mental load of operating the prosthesis [5].

Humans know where their limbs are in space due to proprioceptive feedback cues in the human body. Visually monitoring of the limbs is not necessary to know where the limbs are in space and which forces are acting on the limbs. Compared to externally powered prostheses, body powered prostheses have the advantage of offering direct proprioceptive feedback. The user of a body powered prosthesis can feel the forces and displacements with which he is operating the prosthesis. Up to now, no commercially available arm prosthesis utilizes the full advantage of proprioceptive feedback. Mostly these prostheses require too high operating forces [6, 7]. The high operating forces are assumed to disturb the proprioceptive feedback.

During shoulder controlled prosthesis operation, the user's body movements result in cable displacement, which is directly related to the opening width of the terminal device. The relationships of body movement, cable displacement, and opening width of the terminal device are shown in Figure 1. Since body movements are fed back by the proprioceptive feedback cues to the central nervous system (CNS), the user is aware of his movements. Thus, in a way the user is aware of the opening width of the terminal device without looking at it. Figure 1 also shows that the user's muscle force results in a cable activation force, which is directly related to the pinch force of the prehensor.

The focus of this study is the relationship between the user's muscle force and the cable force. More details of this relationship can be found in Figure 2, which shows an

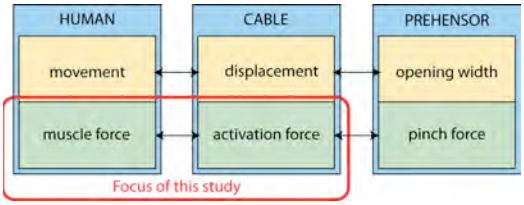

Figure 1: Relationships is prosthesis control

overview of the human-prosthesis control interface. Muscle activation, stimulated by the CNS, results in muscle force. Via the shoulder harness-skin interface and the socket-skin interface the control cable is tensioned, which results in cable activation forces. Since the Bowden cable mechanism causes friction when the inner cable is moving with respect to the outer cable, the cable activation forces are split into cable forces before the Bowden cable, called human cable activation forces, and into cable forces after the Bowden cable, called the prehensor cable activation forces.

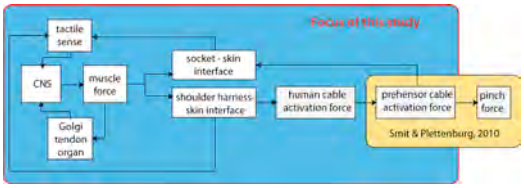

Figure 2: Human-prosthesis control interface

Figure 2 illustrates how the user receives force feedback via his Golgi tendon organs (GTO) and his tactile sense. The GTO sense the created muscle force and transmit this force information to the CNS. Additionally, via the shoulder harness-skin and socket-skin interfaces, the skin senses pressure and sends this kind of force information to the CNS. Because of these feedback paths, the user is aware of his created cable forces. In normal motor control tasks, the GTO

play an important role in force feedback, more than tactile feedback [8].

The quality of the feedback, and thus the performance of the man-machine-system, depends on two components: the mechanical properties of the system, and the window of feedback perception of the human body. This window of feedback perception has a certain range and resolution. Forces and differences in forces can be too low to perceive. Furthermore, a user might notice fluctuations of forces only in a certain frequency range.

The literature does not state at which force levels the human perceives enough feedback to take advantage of the effect of EPP and direct proprioceptive feedback.

The purpose of this experimental research was to find a window of optimal cable operation force, in which a human perceives the best feedback without feeling pain and getting exhausted. Once an optimal operation force window is known, the grasping forces required for daily activities need to be related to the optimal cable forces. This should result in a force transmission ratio for new prosthesis design.

Due to the page limitations of this paper, in the following sections only some of the major issues are discussed. A more detailed description of the experiments and the results will be published in due time.

METHOD

The used measurement procedure was based on the psychophysical measurement method of adjustment [9].

Subjects

Thirteen subjects without arm defects (7 male and 6 female) and 7 subjects with arm defects (4 male and 3 female) participated in this study. Twelve of the 13 subjects without arm defects were right-handed. The subjects of this group were on average 25±3 years old, were 178±10 cm tall, and had a body weight of 71±10 kg. On average, the seven subjects with arm defects were 42±13 years old, were 180±6 cm tall, and had a body weight of 70±7 kg. All subjects with an arm defect were prosthetic users; 3 used a myo-electric prosthesis, and 4 voluntary opening shoulder controlled prosthesis.

Measurement equipment

The hardware used during the experiments consisted of a 'one fits all dummy prosthesis', which was connected to a shoulder harness via a Bowden cable. The Bowden cable was fixated to the 'dummy' prosthesis in such a way that cable displacement was disabled. This setting simulated the grasping of non-deformable objects. A load cell, measuring the cable forces, was connected to the cable, and was located between the shoulder blades of the subject during the experiment. The load cell was connected through an amplifier and a data acquisition system to a laptop, which was running a LabVIEW program. The measurement setup is shown in Figure 3 and Figure 4.

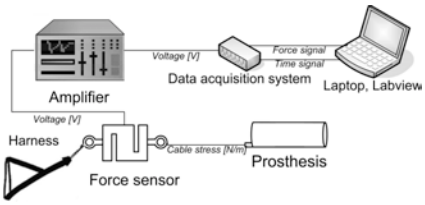

Figure 3: Schematic overview of the measurement equipment

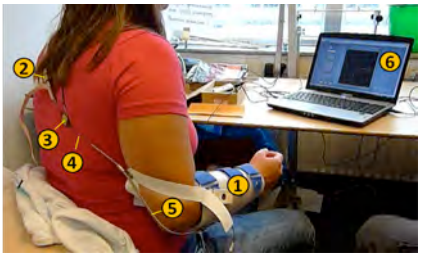

Figure 4: Measurement setup showing the dummy prosthesis (1), shoulder harness (2), force sensor (3), inner Bowden cable (4), outer Bowden cable (5), and laptop with LabVIEW measurement program (6)

Task

Five experiments with five different force levels (5, 10, 20, 30 and 40 N) were carried out. During the experiments, the subject needed to reproduce a given reference force, once while seeing the reference force on the laptop screen, and once without seeing the reference force, Figure 5. The subject was requested to reach the reference force level as fast as possible and hold the reproduced force as constant as possible.

During the experiments the subject wore only a T-shirt, sat on a chair without armrests and looked at the front panel of the LabVIEW program on a computer screen. The dummy prosthesis was placed on the right arm of all subjects without arm defects. Subjects with arm defects wore their own prosthesis. The 'dummy' prosthesis was placed over the prosthesis to establish a better connection with the measurement equipment to the stump. In every other respect, the task was the same for both groups.

The subject was instructed to deliver forces by abduction and adduction of the arm wearing the 'dummy' prosthesis and by protraction of the opposite arm/shoulder or a combination of those three. The subject was free to determine the optimal strategy.

Figure 5 shows the beginning of one experiment at one reference force level. The red line indicates the reference

force, the blue line the reproduced force. The duration of one block was 15 seconds followed by a break of 5 seconds.

A beep identified the beginning and the end of each reference force block wave (Figure 5). Furthermore at every second reference force block wave, the waveform chart was switched off. This means that the subject could see the reference force and the reproduced force at the first reference force block wave (= block with visual feedback) and could not see the reference force and produced force at the following reference force block wave (= block without visual feedback). Because of the beeps the subject knew during the block without visual feedback when to start and stop reproducing the reference force. Another block with visual feedback followed, continued by a block without

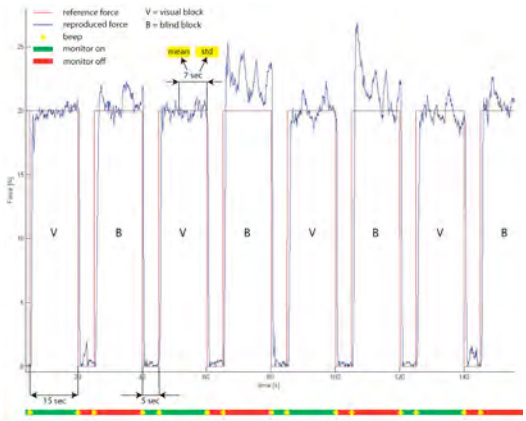

Figure 5: Illustration of the beginning of one experiment at reference force level 20 N. The red lines indicate the reference force, the blue lines the reproduced force. The length of each block is 15 seconds with a break between the blocks of 5 seconds; means and standard deviations (std) are taken from the last 7 seconds of each block.

visual feedback, and so on (Figure 5). One experiment contained 15 blocks with visual feedback (henceforth referred to as visual blocks) and 15 blocks without visual feedback (henceforth referred to as blind blocks).

Five reference force levels (5 N, 10 N, 20 N, 30 N, and 40 N) were measured. The reference force levels were offered to the subject in a randomized order. The vertical axis settings in LabVIEW were chosen in a way that the reference force was always shown in the middle of the vertical axis of the waveform chart.

Performance criteria

The performance of a subject depended first of all on how well a subject was able to estimate and reproduce the given reference force, henceforth referred to as reproducibility. In daily activities we estimate the pinch force our hand creates, which is needed to grasp and hold an object. In an ideal world the pinch force of a prosthesis is directly related to the cable activation force of the prosthesis. A prosthetic user needs to estimate the cable force he is creating using the shoulder harness of the prosthesis. Herewith the estimation of the pinch force of the prosthetic hand is made. A bad estimation of the pinch force might result in the slipping or crushing of a held object.

The mean of the reproduced force of each block was averaged across the last 12 blocks (mean of means). A measure of reproducibility was the deviation of the reproduced force (mean of means) and the reference force.

A second performance criterion was the ability of a subject to hold the reproduced force at a constant level, henceforth referred to as stability. When grasping and holding a vulnerable object, the boundaries of tolerable pinch forces might be narrow. Therefore it is important to be aware of deviating the pinch forces and thus the cable activation force. The standard deviation of the reproduced force of each block was averaged over the last 12 blocks (mean of noise) and was taken as a measure of stability.

Last but not least, a measure of performance was the ability to reproduce the same force several times, henceforth referred to as repeatability. Once a prosthetic user learned the required cable force to grasp and hold a certain object, he needed to be able to recreate this cable force each time he wanted to handle this specific object. Over the last 12 blocks the standard deviation was taken from the mean values of the reproduced force at each block (noise of means). This was taken as a measure of repeatability.

RESULTS

Subjects without arm defect

The difference between the reproduced and reference force was measured the smallest in terms of absolute and relative reproducibility between the 20 and 30 N experiments. The absolute stability and repeatability minima are found at 5 N, whereas the minima for relative stability and repeatability are found during the 30 N experiments.

The higher the reference force level becomes during the experiments, the higher the average values of absolute stability and repeatability become. Additionally, the higher the reference force level becomes during the experiments, the higher the deviation across the group of subjects becomes for reproducibility and repeatability.

During the 5 and 10 N experiments, the highest deviation between the reproduced and reference force result in terms of relative reproducibility. Additionally, the highest values in terms of relative stability and repeatability are found during these experiments. Furthermore, the deviations between the

subjects' results are the highest for the 5 and 10 N experiments in terms of relative reproducibility, stability and repeatability.

SUBJECTS WITH ARM DEFECTS

All seven subjects with arm defects succeeded in finishing the 5 N experiment, whereas only six of the 7 subjects were able to complete the 10 and 20 N experiments. The 30 N experiment was carried out by four of the seven subjects with arm defects and three of the tested seven subjects succeeded in carrying out the 40 N experiment.

The results of the subjects with arm defects were compared with the subjects without arm defects. For the subjects with arm defects, the force where the reproduced force equals the reference force is found between 10 and 20 N in absolute and relative sense, whereas these values are in between 20 and 30 N for the subjects without arm defects. The results for stability and for repeatability of both groups overlap in absolute and relative sense.

CONCLUDING REMARKS

The purpose of this study was to find an optimal operation force, at which the prosthetic user receives the best force feedback during comfortable prosthesis operation. Three performance factors were introduced: reproducibility, stability, and repeatability. The following conclusions can be made about the subjects without arm defects:

- An optimum is found between the 20 and 30 N experiments for absolute & relative reproducibility.
- The optimum for relative stability and repeatability is found during the 30 N experiment.
- The optimum for absolute stability and repeatability is found during the 5 N experiment.

Although the optima for absolute stability and repeatability are found for the 5 and 10 N experiments, these operation forces cannot be called the optimum, as these experiments show the worst performance and highest deviation across a group of subjects in terms of relative reproducibility, stability and repeatability.

The 40 N experiments show no significant differences to the 30 N experiments, where the optimum operation force is found, in terms of absolute and relative reproducibility as well as for relative stability and repeatability. Still, an operation force of 40 N cannot be called an optimum because the highest deviation between the subjects is found in terms of absolute reproducibility at this force level. Thus, subjects are not always equally capable of reproducing a certain force. Moreover, the results of the 40 N experiments show the worst performance and highest deviation between subjects in terms of absolute stability and repeatability. This means that subjects are not capable holding a force at a constant level during one block and have difficulty reproducing the same force at different moments in time.

Another objective of this research is based on the question: Can the performance of a person with arm defect be predicted using the experimental results of subjects without arm defects? The stability and repeatability performance of subjects with arm defects does not differ to the performance of subjects without arm defects. The same is found for the absolute and relative reproducibility of the 5 and 10 N experiments. However, for the 20, 30 and 40 N experiments, the average results across the group of subjects with arm defects are lower than the lower standard deviation border of subjects without arm defects for absolute and relative reproducibility. Therefore, a difference in reproducibility performance is found for the three higher forces between subjects with and without arm defects. Furthermore, the reproduced force equals the reference force between 10 and 20 N for the subjects with arm defects, whereas this optimum is found between 20 and 30 N for subjects without arm defects. Thus, another difference in reproducibility performance is found between subjects with and subjects without arm defects.

The fact that subjects with arm defects did not succeed in performing experiments with the higher reference forces (30 and 40 N experiments) implies that those cable forces are too high to operate during daily activities for subjects with arm defects. Indeed, the lower optimal force level for reproducibility performance emphasizes this conclusion.

In summary, the following points can be concluded from this research:

- The optimal operation force, at which the user receives optimal feedback and is able to control the prosthesis comfortably, is found between 20 and 30 N for subjects without arm defects.
- A lower optimal operation force between 10 and 20 N is found for subjects with arm defects. Stability and repeatability performances of subjects with and without arm defects are comparable.
- Cable forces between 5 and 10 N are too low to be controlled with optimal force feedback.
- The border of comfortable operation is found around the cable activation force of 40 N. At this boundary, the proprioceptive feedback is disturbed.
- A uniform prosthesis design can be used for female and male prosthetic users.

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CONTINUOUS POSITION AND FORCE CONTROL OF A MULTIGRASP MYOELECTRIC TRANSRADIAL PROSTHESIS

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INTRODUCTION

Dependable and efficient utilization of a multigrasp prosthetic hand requires an effective control interface. This interface should be intuitive and direct, offering continuous and proportional control of motion with negligible latency. Realization of such a controller is a challenging problem in upper extremity prosthetics research although several significant strides have been made. Prevalent approaches to multigrasp control thus far include pattern recognition [1-6] and hierarchical control [7-11].

This paper presents the design and preliminary experimental validation of a myoelectric controller that is intended to control the continuous motion of a multigrasp prosthetic hand between nine characteristic postures (reposition, point, hook, lateral pinch, opposition, tip, cylindrical, spherical and tripod). The controller, referred to as multigrasp myoelectric control (MMC) is based on an EMG supervised event-driven finite state machine. The EMG component provides user intent, and consists of a single bipolar signal acquired through two EMG electrodes, similar to EMG interfaces commonly found in commercial myoelectric prostheses. The state machine acts in conjunction with a low-level coordination controller to activate different actuator subsets (connected to digits via tendons in the prosthesis) based on the present state. The controller incorporates object detection and force estimation algorithms to allow force based state transitions and the estimation of digit forces.

To test the functionality of the controller, experiments were conducted on a healthy subject using an able bodied adapter with a multigrasp prosthetic hand. Experimental results are presented that demonstrate the ability of the MMC to provide effective movement and grasp control of the multigrasp prosthesis.

MULTIGRASP MYOELECTRIC CONTROL

The MMC consists primarily of a uniquely structured finite state machine (see Fig.1) and a coordination controller. The output of the state machine, the current hand state (posture), dictates which subset of actuators (and associated tendons) are active in the hand at any given time. The active tendons are indicated on the inset of each state in Fig.1, where T1 controls Digit II Flexion, T2 controls digits III-V

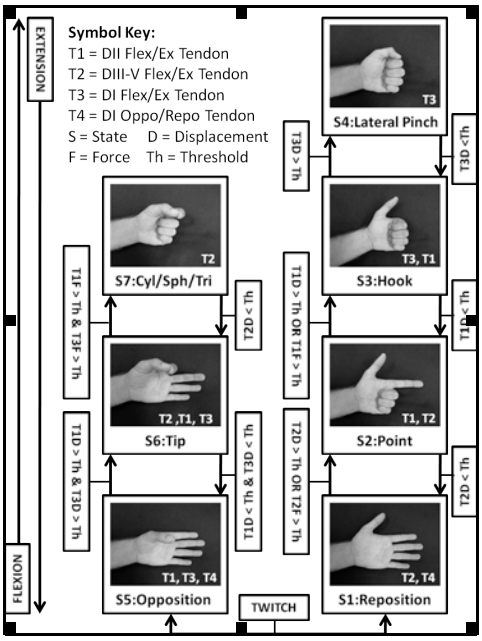

Figure 1: Structure of the MMC state-machine

flexion simultaneously, T3 controls digit I flexion, and T4 controls digit I opposition. The position references for these actuators are driven by proportional signals arising from the EMG input. Changes in digit position or digit grasping force trigger transitions in the state chart based on pre-established thresholds. Twitch commands (a high intensity co-contraction of the muscles at both electrode sites) may also cause transitions among the reposition (platform) and opposition postures. Once a transition occurs, the current state of the hand changes, and a new subset of actuators and associated tendons become activated by the coordination controller. The active actuators are associated with transitions to adjacent states. This configuration is intended to leverage the benefits of traditional myoelectric control by allowing for the direct and proportional control of motion of a multigrasp hand from a single EMG input (i.e., one pair of EMG channels). A more detailed explanation of this controller may be found in [12].

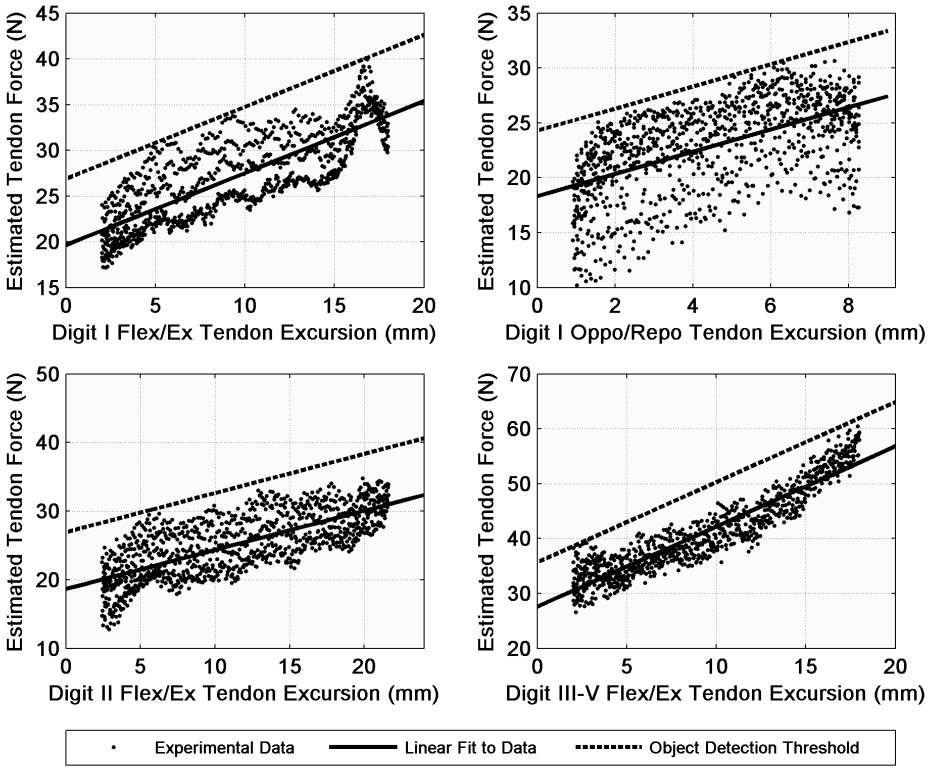

Figure 2: Estimated Tendon Force versus Tendon Excursion for Flexion of Digits I-V and Opposition of Digit I

OBJECT DETECTION AND FORCE ESTIMATION

Object detection and force estimation were implemented in the MMC to enable force-based transitions in the state chart and allow for proportional force control. To do this, the digits of the hand described in [13] were driven repeatedly through their full range of motion with a chirp signal whose frequency increased exponentially from 0 Hz to the motion bandwidth of each digit (or joint, in the case of the digit I opposition degree of freedom). The current command and tendon excursion were recorded during these motions. The tendon force, F_T, was then estimated as $F_{T=\frac{u_m k_t N}{r}} = \frac{u_m k_t N_G}{r}$, where u_m is the motor current, k_t is the motor torque constant, N_G is the gearhead ratio, and r is the pulley diameter of the hand described in [13]. The graphs in Fig. 2 depict the force required to either flex the digits or oppose the thumb as a function of tendon excursion. A linear fit was then applied to these data (ignoring the first and last 10% range of motion) and offset by the maximum difference between the experimental data and the linear fit. Note that, by using the chirp signal to generate this data, dynamic effects due to variations in velocity (i.e. friction) and acceleration (i.e. inertia) are accounted for. This being said, the spread of the data for a given excursion is usually on the order of 10 N. As this represents at most approximately 4% of maximum tendon force (270 N for short-term operation) this variation is assumed to be insignificant, and a quasi-static characterization may have been sufficient. Nevertheless, this process established a conservative characteristic baseline for unimpeded motion which was utilized as an object detection threshold for each degree of actuation, respectively. An object was detected when the instantaneous tendon force estimate during operation (based on the above equation and dependent on motor current) exceeded the object detection threshold (dependent on tendon excursion).

A proportional signal was generated by subtracting the instantaneous tendon force estimate from the object detection threshold. The normalized finger force was then found by dividing this quantity by the maximum force achievable given the thermally induced current limits of the motors.

EXPERIMENTAL PROCEDURE

To test the MMC with object detection and force estimation, the multigrasp prosthesis described in [13] was attached to a healthy subject using a custom built, able-bodied-adapter, depicted with the prosthesis in Fig. 3.

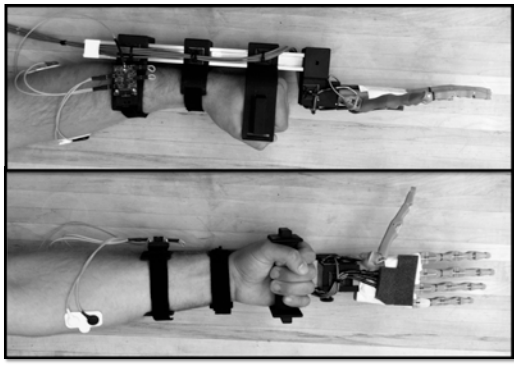

Figure 3: Prosthesis and able bodied adapter

To verify the efficacy of the object detection algorithms, the subject was required to traverse the state chart while grabbing various objects to impose both position and force based state transitions. Specifically, a roll of electrical tape was grasped while in the point state to impede motion of digit I, and a 6 cm (2-3/8 inch) diameter PVC pipe was grasped while in the tip state to impede motion of digits III-V (see Fig. 4).

RESULTS AND DISCUSSION

Figure 5 shows EMG control input, hand state, tendon excursion, and normalized finger force during the experiment. This figure demonstrates several important characteristics of the MMC. First, the same EMG input can affect positional references for different actuators based on the current state of the hand (EMG channel 1 commands T2 around the reposition state and controls T3 around the lateral pinch state). Second, a single EMG input may govern multiple actuators, (EMG channel 1 simultaneously controls actuators T1 and T3 in the opposition and tip states). Third, a high intensity co-contraction of the forearm flexor and extensor muscles results in a twitch. The twitch event causes automated opposition and reposition of the thumb (note the behavior of T4 after the occurrence of a twitch). As can also be seen in Fig. 5, response to user intent is immediate. That is, movement occurs as soon as elevated EMG signal levels are detected.

Figure 5 also verifies that force-based transitions were successfully executed as indicated in the figure by arrows. It can be seen that a force-based state transition occurred between the point and hook states when the estimated tendon force for T1 exceeded the object detection threshold as the electrical tape was grasped. This transition then allowed T3 to flex and further enclose the grasped object. Similarly, a force-based transition occurred between the tip and cylinder/spherical/tripod grasps as tendons 1 and 3 began to close around the 6 cm (2-3/8 inch) diameter PVC pipe. Although these fingers were able to close sufficiently to cause a transition to the tip state, the occurrence of object detection allowed further transition to the cylinder/sphere/tripod grasp state. This, in turn, allowed T2 to flex, causing digits III-V to close, and adding further stability to the grasp. While previous work [12] had demonstrated the efficacy of this controller in a virtual environment, with tendon excursion based transitions only, this was the first demonstration that the controller was effective with hardware, and that force based transitions could be executed successfully in the presence of grasped objects. Finally, Fig. 5 also shows that the normalized finger force increases with continued EMG input after object detection has occurred, providing a signal which may be utilized for user feedback.

CONCLUSION

This paper demonstrates that the MMC provides direct access to multiple grasps and postures with negligible latency. By grasping a variety of objects while traversing the state chart, it was seen that the object detection and force estimation

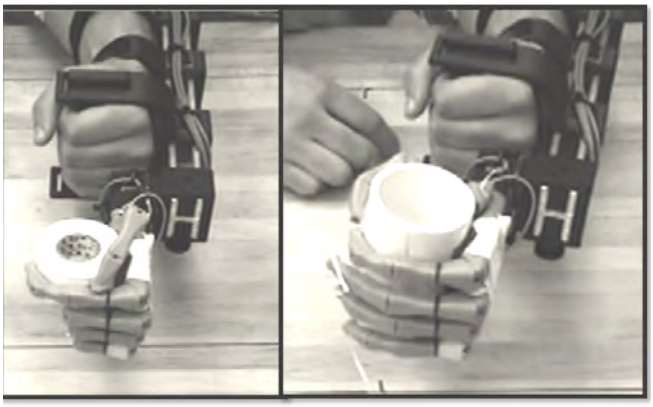

Figure 4: Objects grasped during experimentation

algorithms are functional and allow for continuous force and position control. This was the first physical (as opposed to virtual) demonstration of the controller's effectiveness. In future work, the MMC and multigrasp prosthesis will be functionally assessed on amputee subjects. Additionally, normalized finger force will be utilized to provide some form of feedback to the user.

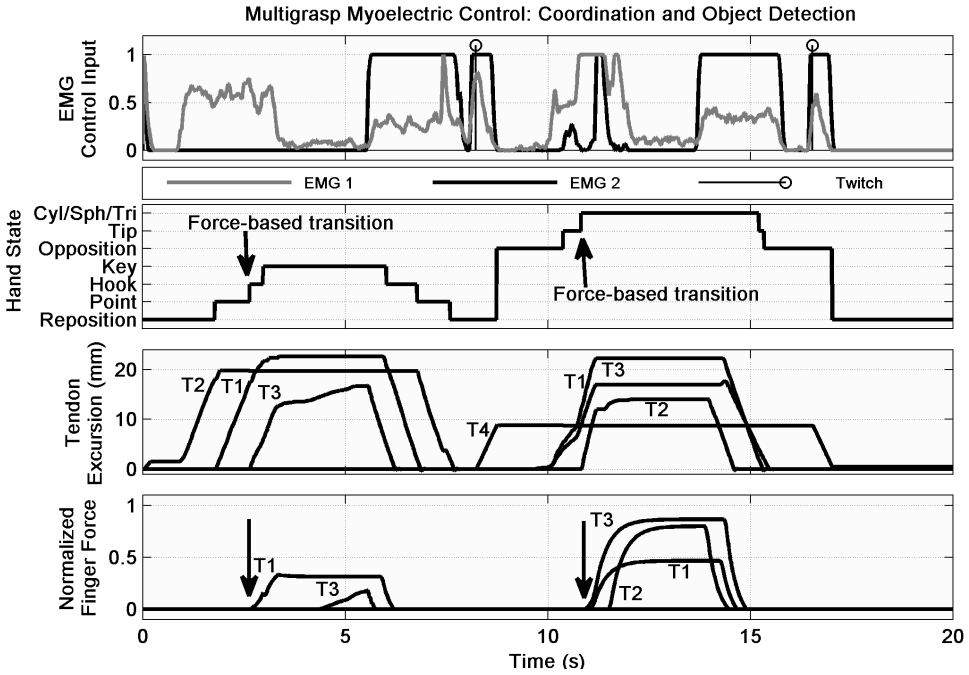

Figure 5: EMG input, hand state, tendon excursion, and normalized finger force during state chart navigation

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THE DESIGN OF A MYOELECTRICALLY CONTROLLED HAND WITH MULTIPLE ACTUATORS FOR FIVE-YEAR OLD CHILDREN

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ABSTRACT

Myoelectric prosthetics are complex functional devices that can improve significantly a person's quality of life. This paper describes the development of a myoelectrically controlled prosthetic hand for a five-year old child. A key consideration in the design of upper-body prostheses is to use information from studies highlighting the main causes of rejection. These studies emphasize that in order to reduce rejection, it is necessary to include the opinions of the users in the design process. Additional constraints are introduced due to the small size and mass of a five-year old child's hand compared to that of an adult. The main points of the final design are detailed, including the areas where these constraints were overcome. Modularity was used throughout the design; it allows the hand to be configured for the individual user, and also helps to reduce the potential cost of the hand. The final design has three actuators controlled individually through the use of a master-slave microchip combination. This design has a final mass of 105.8g and produces a pinching force of 4.35 N.

INTRODUCTION

There have been greater advances in the design of prosthetic hands for adults compared to those for children. Although there have been developments to child prostheses, they have not always been in line with those made to adult prostheses. Acceptance of the user is a key consideration in the design of upper-body prosthetics. It is generally recognised that the younger a user is introduced to a myoelectrically controlled prosthesis, the greater their acceptance of the technology [1]; this is encouraging the fitment of functional and adaptable prosthetic limbs to young children. To provide choice, hands designed specifically for the needs of children are required. Currently there are two commercially available upper-limb prostheses specifically designed for children: the Otto Bock 2000 Electric Hand, and the RSL Steeper Scamp Myo Electric Hand. Both of these hands are single degrees of freedom devices that are available in various sizes, and driven by a single actuator that closes the first and second fingers onto the thumb. Improvements in child prosthetics could be made with improved adaptability and an increased number of individually driven axes. To address this, the development of prostheses for children that are produced in conjunction with research into the acceptance and needs of children is needed. This paper describes how a prostheses for young children was designed with multiple degrees of freedom, modularity and functionality, taking into account considerations from both a user's perspective and from technical constraints. (A final prototype can be seen in figure 1.)

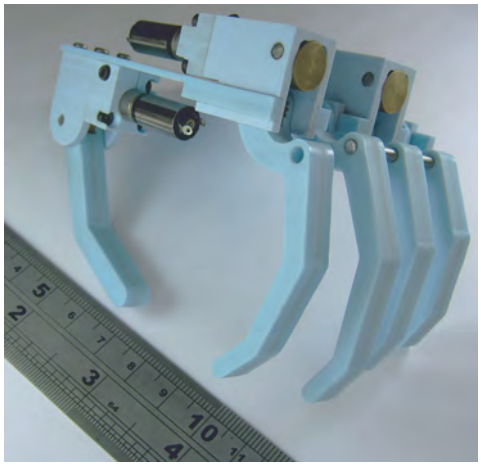

Figure 1 – A Prototype Myoelectric Hand.

USER CONSIDERATIONS

Rejection rates of upper limb prostheses amongst children have been reported to be as high as 50% [2]; indicating that upper limb prostheses that are currently being prescribed are not meeting the needs of young people [3]. Research into rejection of prostheses amongst adult users found dissatisfaction with the prosthesis to be linked to rejection [4], therefore highlights the importance of including the views of users when developing new prosthetic devices. This is supported by Bidiss & Chau's [3] historical review of upper limb prosthetic use and abandonment, which concluded that "increased emphasis on participatory research and consumer satisfaction is needed"

Bidiss et al [5] involved prosthetic wearers of all ages to inform prosthetic design by identifying their key development

priorities. These were reduced weight, lower cost, life-like appearance, improved comfort, enhanced wrist movement and better grip control/strength. The design priorities varied substantially across age groups, suggesting that upper limb prostheses designed from the users' perspective would be different for children compared to those designed for an adult. This supports the need for prosthetic hands for children designed alongside studies into the views of the users. Before this user-led design, it is necessary to explore the technical feasibility of designing a hand of this size and mass.

At Southampton University a study (Our Bodies Our Views) used questionnaires and interviews to examine satisfaction with prostheses and reasons for prosthesis rejection in young people with upper limb loss aged 5-18 years. Three factors were identified as important amongst the participants. They were: the look of the prosthesis; the functional ability, and being involved in the selection of the prosthesis. Reasons identified for not wearing the prosthesis were: it was uncomfortable (including being too hot and too heavy); that it is only useful for specific tasks; the artificial appearance of the prosthesis (attracting unwanted attention), and wear and staining. This study also highlighted the importance of communicating with children when designing prosthetic devices.

TECHNICAL DESIGN CONSTRAINTS

			5yr	16yr	% Dif
A	Hand length	mm	125	187	66.8
B	Middle finger length	mm	52.5	80	65.6
C	Palm length	mm	72	107	67.3
D	Palm width	mm	57	82.5	69.1
E	Ratio of palm length to middle finger length	%	42.35	42.75	99.1
F	Ratio of palm width and length	%	82.5	80	103.1

Table 1: Hand Measurements of 5 and 16 Year Olds [6].

When designing prostheses for children there are issues introduced due to the differing size and mass requirements. Table 1, for example, shows average hand measurements for 5 and 16 year olds [6]. The data in rows E & F, shows that irrespective of age, certain proportions of the hand are virtually unchanged. However, the natural hand of a five year old child is two thirds smaller than that of the average 16 year old (approximately equivalent to an adults hand); suggesting a similar difference in the overall mass. The effect of this constraint is most prevalent in the design of the drive system, where the consideration of output power and speed are equally important. However larger actuators are typically heavier. Including multiple functional axes means that multiple drive systems are required; as a result there is a summing effect of the significance of the drive system weight.

DESIGN OF A PROTOTYPE HAND

To realise a design that is both cheap and flexible, the decision was made to include a high level of modularity. This would be split into two levels. The first level would be in the manufacture to aid in reducing the number of different parts and construction processes, therefore, reducing the cost of manufacture. The second is to provide technician level reconfiguration; to provide the user with flexibility and choice when choosing their exact specification. This permits easy setup, reconfiguration and maintenance of the hand; possibly allowing for reduced post-fitment costs.

An electric motor and gearbox was used to actuate the hand since it is the common method of actuating myoelectric prosthetic hands. The design of the gearbox arrangement is based on a scaled version of the Southampton Hand's gearbox [7]. It uses Faulhaber DC-Micromotors (0816 with a 64:1 gearbox) to drive the fingers and thumb through a worm-wheel combination. The defining characteristics of a drive system are the output speed and torque. Both of these values are determined by the characteristics of the motor and gear chain. Equation B (Appendix A) shows that the gears have a linear effect on the output torque and an inverse relationship with the output speed.

The motor selected for this project produces 0.15 mNm and rotates at 15,800 rpm (263.3 rps). There are two gear combinations in the drive chain, the first has a ratio of 64:1 and the second has a ratio of 20:1, with respective efficiencies of 60% and 89%. The torque across a gear system increases proportionally by the ratio of the number of teeth on the gears in the system, the speed through the system decreases with the same relationship. This determines the output characteristics, of 0.12 N maximum force and a maximum speed of 0.13 rps.

Two essential considerations were identified for the design of the prosthesis: the speed for 90º closure of the hand and the force produced at the fingertip. It is assumed that the fingers only rotate through 90º.

Equations C and D were used to convert the drive system output characteristics into prosthetic output characteristics. Equation C gives a closure time of 1.95 s. Equation D shows that to calculate the force at the fingertip, the length of the finger from the rotating axis is needed. This design has a middle finger measuring 55 mm which gives an output force of 2.17 N. This produces a theoretical combined finger closure force of 4.35 N. These characteristics are not optimal

and improvements do need to be made in the speed and force generation. However, it was decided since the hand was for a preliminary study these characteristics would be acceptable.

The artificial metacarpophalangeal (MCP) joint is defined as the key component in the design, as it houses all of the driven components of the hand. As mentioned previously, the design is based on the Southampton Hand [7]. However scaling the design needed careful consideration to ensure adequate strength of the components. The design incorporates the axle for the motor and is split to allow the worm to be placed into the MCP joint. A key feature in this design is the connection slot to allow the MCP to fit into any of the four MCP locations on the palm.

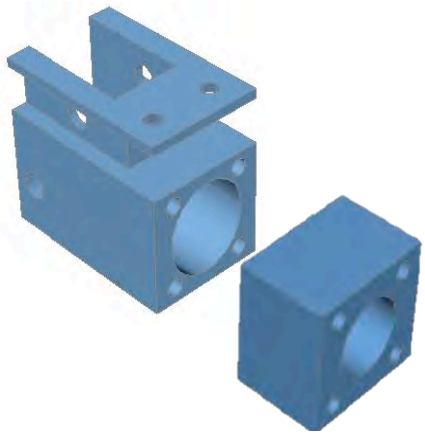

Figure 2 - Prosthetic Metacarpophalangeal (MCP) Joint

The shape of the fingers and thumb was chosen to mimic that of a human hand and to allow the first finger and the thumb to form an effective pinch. The base of the finger has a slot to allow for a strong and effective coupling to the wheel gear. The curved base of the finger is aligned with the MCP joint when straight; this allows the fingers to lie flat when fully extended.

The hand uses a microchip-based control system in a master and slave configuration. This design increases the modularity of the system; allowing for easy reconfiguration and motor addition. It uses an overcurrent device to regulate the force at the fingertips but has the availability to incorporate embedded force sensors into the fingertips. The current system though functional, does not provide closed feedback required for fine touch.

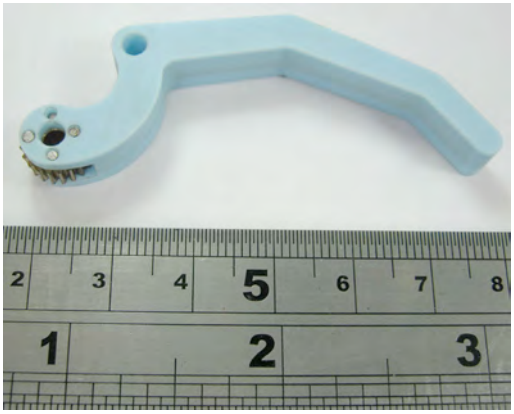

Figure 3: A Prosthetic First Finger.

DISCUSSION

This study shows that it is possible to build a prosthetic hand that incorporates multiple actuators for children aged five-years. The final prototype is 127 mm long and 60 mm wide; these values are comparable to the size of a five-year old human hand. The mass of this design is 105.8g; this value is similar to that of existing prosthetic hands for children. However, the mass can be reduced through material changes and design alterations. All of the components of the drive are interchangeable throughout the system; including the motors, gears and all drive shafts. The hand has only 22 different mechanical parts; including 7 drive shafts, screws and pins that all require minimal manufacturing. The second level of modularity allows for the hand to be reconfigured to fulfil the exact requirements of individual users without any adjustment to the design. An example of this is that the middle finger for one user may be the index finger for another. This would reduce the total amount of components that a fitment centre stocked, therefore, potentially reducing the costs.

CONCLUSION

This novel, child prosthetic hand is fully adaptable, whilst, still providing a high level of functionality. The design confirms that it is feasible to provide hands for children that are able to deliver choice, without compromising on the size or mass. The power of the drive system may be increased without affecting the target age and functionality and can be achieved by changing the motor and the design of the MCP joint. The modularity in the design added significant functionality and showed that it could increase the choice given to the users, whilst reducing pre- and post-fitment costs. This area of research calls for further development.

FUTURE WORK

This study highlights several areas for possible improvements, the first of which would be to increase the speed and force characteristics. Further studies will be undertaken to improve the control system by including force and position sensors allowing for the development of a hybrid force-position control system. This could be implemented with the use of encoders on the motor shafts to infer position of the fingers. During a redesign, the mass of the hand could be reduced further with the use of different materials and an altered drive system. The modularity incorporated into the design could be adapted to provide in-service reconfiguration. This would further increase the functionality and could reduce the need to service the entire hand.

Having confirmed the feasibility of producing a hand with suitable size and mass characteristics, research focusing on the users' views is needed. Although this study begins to address user considerations and reasons for rejection this was not extensive. Therefore further research will be conducted to investigate the aspects of prostheses that are important to children, and to explore their views on new designs for future devices.

ACKNOWLEDGEMENTS

The authors of this paper would like to thank Mark Long and ECS Zepler Mechanical workshop M.S. Abd Rahman, A.M. Abdullah, A.G. and M.H. Mat Saman for their advice and work in the manufacture of the hand.

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APPENDIX A

Gear ratio equations:

$$\text{gr} = \frac{(\text{Teeth on Input})}{(\text{Teeth on Output})} = \frac{n_{\text{in}}}{n_{\text{out}}} \quad \text{(A)}$$

$$gr = \frac{n_{\text{in}}}{n_{\text{out}}} = \frac{\tau_{in}}{\tau_{out}} = \frac{\omega_{out}}{\omega_{in}} \quad \text{(B)}$$

Where,

n_{in} or n_{out} = Number of teeth on input or output shaft

τ_{in} or τ_{out} = Torque on input or output shaft

ω_{in} or ω_{out} = Rotational velocity of the input or output shaft

APPENDIX B

Time for 90º rotation:

$$t_{90} = \frac{1}{4\omega_{out}} \quad \text{(C)}$$

$$t_{90} = \frac{1}{4 \times 0.13} = 1.95s$$

Where,

t_{90} = Time for 90º rotation

ω_{out} = Drive shaft rotational velocity

APPENDIX C

Equation of moments:

$$F = \frac{\tau}{r_f} \quad \text{(D)}$$

$$F = \frac{0.1}{5.5 \times 10^{-2}} = 1.86N$$

Where,

r_f = length of finger from rotating axis.

τ = Torque

F = Force

A ONE YEAR RETROSPECTIVE OVERVIEW OF PARTIAL HAND PATIENTS USING PRODIGITS

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ABSTRACT

To date, more than 150 patients worldwide, with partial hand amputations, have been fit with ProDigits technology. This study includes a comprehensive overview of 14 of those patients. A major emphasis will be placed upon individuals with partial hand loss due to congenital limb deficiency versus traumatic/disease partial hand loss, and individuals with unilateral and bilateral partial hand amputations. Four different domains are included in this study that represent how these 14 individuals perceive their "Improved Self Image," "Increased Independence," "Positive Change in Lifestyle" and "Increased Activity and Participation in Daily Life." In order to adequately measure these parameters, an overview of a Client Centered Care System (3CS) assessment will be demonstrated, as it presents a new evidence-based tool for upper limb amputee outcomes measurement.

INTRODUCTION

It is stated by Hill et al in "Upper Limb Prosthetic Outcome Measures (ULPOM) : A Working Group and Their Findings," that advances in the design, control, application and provision of upper limb prostheses in recent years has required a more objective justification for the costs involved in providing these services. This has intensified interest in objective measures of performance and use of artificial arms. (1) Without a more unified approach to define what constitutes true "success" in upper limb prosthetic utilization, we cannot effectively communicate between professions, rehabilitation centers and countries. As technology advances the methods of measuring outcomes and patient success must also advance.

A State of the Science Conference (SSC), to address this need, was convened in Chicago, Illinois, in March 2009. The goals of the meeting were to examine the body of scientific knowledge that related to outcome measures in upper limb prosthetics and to examine the following: validated instruments to measure upper limb prosthetic outcomes, what do these instruments measure as it relates to the International Classification of Functioning, Disability and Health (ICF)/ World Health Organization (WHO) classifications, strengths and weaknesses of current instruments, appropriate tools for various applications and primary future research priorities. (2)

After examining this extensive body of knowledge from the SSC, a Client Centered Care System (3CS) was developed by Diane Atkins, OTR, Karl Lindborg, CPO, and a research team including an independent MD and 3 PhD researchers from the Matrix Health Center, LLC. This project began in January 2010 and it continues to be a work-in-progress. A primary goal of this research was to create a comprehensive, multi-disciplinary patient care process that was client-oriented and optimized positive outcomes. Accurately assessing the client's needs and establishing realistic expectations that align the capabilities of the client with those of the device, and experience occupational therapy training with the device, are keys to maximizing the retention rate. Client management begins with a "Candidate Review process" and is designed to flow seamlessly throughout the continuum of care in order to optimize outcomes and client satisfaction. The Continuous Quality Improvement (CQI) of the 3CS process is designed to help ensure effective, efficient and timely ongoing feedback to optimize the entire system. This, in turn, provides additional support to maximize patient success, and thus improve the retention rate.

An additional major goal of this effort was to create a series of assessment tools that gathered data as it related to the important parameters of function (device performance and client satisfaction with the device), independence, general health, activity level, pain, overuse syndromes, occupation, leisure/recreation, social adjustment, self-image, goal setting, motivation, resiliency and quality of life (QoL).

This Client Centered Care System assessment scoring, interpretation and analysis is designed for clinical use and further studies in order to support the facilitation of improved outcomes and optimize patient/client satisfaction.

METHOD

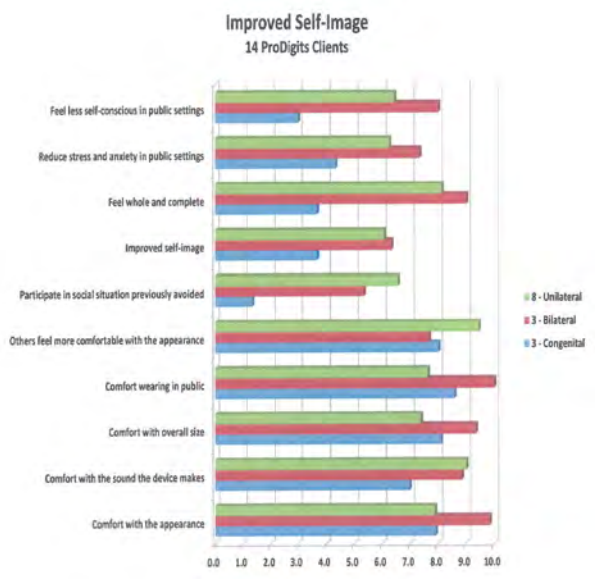
A series of assessment tools were administered to 14 individuals, with partial hand loss, at the time of their initial

evaluation and at various follow-up intervals. The minimum amount of time that an individual was wearing a ProDigits prosthesis for this study was 3 months.

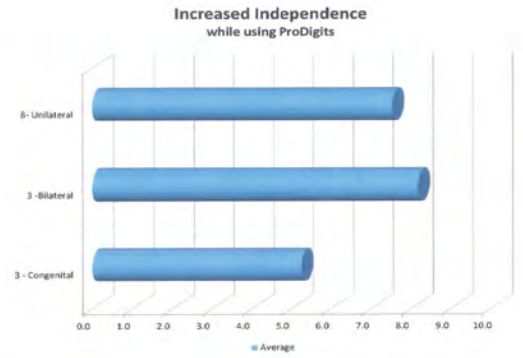
For the purposes of this preliminary study, and of this limited number of ProDigits users, focus was placed upon a 6 page Follow-up Assessment instrument, with over 100 data points, that captured numerous domains. The emphasis of this paper will focus upon 4 areas and the perceptions of the ProDigits users as it related to: "Improved Self-Image", "Increased Independence", "Positive Change in Lifestyle" and "Increased Activity and Participation in Daily Life."

RESULTS

The following 4 graphs demonstrate some of the findings of this comprehensive Follow-up Assessment tool as it relates to 8 Unilateral ProDigits users, 3 Bilateral ProDigits users and 3 Unilateral Congenital ProDigits users (who were viewed separate and apart from the individuals who had sustained traumatic partial hand loss).

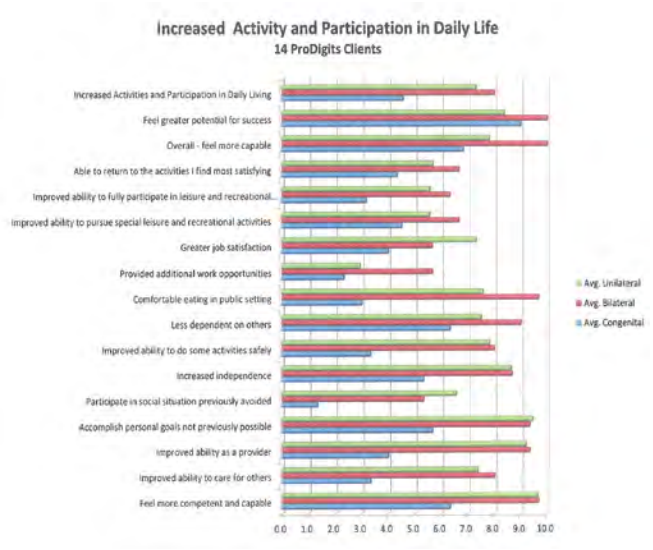


This assessment yields the most dramatic results for the 3 individuals with bilateral partial hand loss. This was the most obvious in the areas of "comfort wearing in public" (9.8/10) and "comfort with the appearance" (9.6/10). Those with unilateral partial hand loss expressed the most positive response as it related to "others feel more comfortable with the appearance" (9.3/10). The 3 individuals with unilateral congenital partial hand absence expressed the highest response in "comfort wearing in public" (8.5/10).



The 3 individuals with bilateral partial hand loss expressed the highest level of perceived "increased independence" (8.9/10), when compared to the 8 individuals with unilateral partial hand loss (7.2/10). The 3 individuals with unilateral congenital limb absence felt that ProDigits increased their independence at a level of 5.0/10.

This "spider web" chart demonstrates the parameters of well-being, appearance, independence, social, occupation, leisure and recreation, as well as quality of life measures. It is apparent that individuals with bilateral and unilateral partial hand absence feel a greater degree of "positive change" as it relates to the aforementioned domains, when compared to those with unilateral congenital limb absence. The area of greatest improvement, as it relates to a "Positive Change in Lifestyle" was in "well-being" with a 9.2/10 overall score for the 3 individuals with bilateral partial hand loss, 9.0/10 for individuals with unilateral partial hand loss, and 5.8/10 for those with congenital partial hand absence.



Individuals with bilateral partial hand loss were favorably impacted in the following areas: "Increased activities and participation in daily living," "Overall–feel more capable," "Improved ability to fully participate in leisure and recreational activities," "Comfortable eating in public settings" and "Improved ability as provider." Individuals with unilateral partial hand loss expressed the most significant improvements in; "Greater job satisfaction" and "Accomplish goals not previously possible." Those with congenital limb absence felt improved capabilities, but not to the degree as those with unilateral and bilateral limb loss. The one area where individuals with congenital limb absence clearly felt an improvement was in "Feel greater potential for success" (8.9/10).

DISCUSSION

The results generated from these analyses are enlightening and informative. The 3 distinctions of the groups studied, Congenital, Unilateral and Bilateral enables the clinician to better visualize the unique differences between those who have been born without part of their hand and those who have lost part of their hand secondary to traumatic injury or disease. The individual who is born without part of their hand experiences life in an entirely different manner. In their responses, all of these individuals viewed themselves at "baseline" as completely independent. They simply learned from childhood to accomplish tasks in a different manner. It is interesting to note that in spite of their perceived independence prior to receiving ProDigits, they indeed see benefits and value in ProDigits as it related to "increased independence" and "improved self-image".

The eleven individuals who had lost part of one or both hands in traumatic injury, or disease, had similar objective responses particularly in the areas of "overall well-being" and "independence." These reactions were verified in the many subjective responses expressing; "It gave me back my confidence in a way that I can live going forward," "I can now shake hands, as I did before, with people looking at *me*, and not part of my hand." The final chart validates all of these findings, and more, in a comprehensive manner. Those with congenital limb absence found value and benefit with ProDigits particularly as it related to "Feeling greater potential for success" and "Overall feeling more capable." Those with bilateral partial hand loss were impacted most by ProDigits as it related to "Increased activity and participation in life."

CONCLUDING REMARKS

This analysis serves as merely an initial "snapshot view" of the many "arenas" of interpretation an assessment tool such as this provides. While this data is interesting and informative, it is extremely preliminary as it only includes 14 individuals with partial hand loss. Additional research, using a larger number of subjects, will provide supportive information for the trends that this study illustrates.

Much of this information is of no surprise to experienced clinicians in this field. This is a "first" however, to validate these findings, and confirm our beliefs, in an evidence-based manner, as it relates to the person with partial hand loss.

The intent of the researchers involved with this project, and this assessment tool, is to continue to test, refine and define this evidence-based, client-centered care system (3CS). As new prosthetic components become available, we want to explore opportunities to utilize these instruments for comparative studies with other upper limb prosthetic devices. Our goals include continuing to document patient data in an objective manner at base line, exit from training, and scheduled follow-up intervals in order to; measure patient results in a comprehensive and quantitative manner, measure patient care process, and improve the prosthetic device itself as objective feedback is provided.

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Wednesday, 17 August 2011

Presentation of Posters - 3:00 pm - Foyer of the Wu Conference Centre

Poster Title	Presenter
A Comparison Study of EMG Features for Force Prediction	A. Smidstrup
Towards an Optimal Model for the Estimation of Force from Intramuscular EMG	J.C. Rosenvang
Resolving the Limb Position Effect	Anders Fougner
Electrically Conductive Silicone Interface for Myoelectric Prostheses with Silicone Socket Suspension	Ian Whatmough
Coding Scheme for Characterising Gaze Behaviour of Prosthetic Use	Mohammad Sobuh
A New Voluntary Closing Hook Prosthesis	Dick H. Plettenburg
Voluntary Closing Prostheses - An Answer to User Needs!	Dick H. Plettenburg
Handy Hook Revisited	Wayne Daly
Towards Optimizing the Electrode Configurations to Improve Myoelectric Pattern Recognition	Aaron J. Young
A Novel Research and Clinical Approach to Using Gel Liners for Collection of Surface EMG Data for Myoelectric Control	Robert Lipschutz
Virtual Reality Simulator for Training and Evaluating Myoelectric Users	Joris M. Lambrecht

A COMPARISON STUDY OF EMG FEATURES FOR FORCE PREDICTION

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ABSTRACT

Myoelectric prosthetic devices can be controlled by use of surface electromyography (sEMG). However, intramuscular EMG (iEMG) has been proposed as an alternative, since it may provide more stable and selective recordings with several advantages. The purpose of this study was to assess the predictive capabilities of 14 features of iEMG and sEMG for force ranging from 0 to 100 % maximum voluntary contraction (MVC). Intramuscular EMG and surface EMG were recorded concurrently from the muscle flexor digitorum profundus from 11 subjects who exerted four force profiles during power grasping. The predictive capability of each feature was assessed using the mean R^2-value with a 1st order polynomial (linear prediction). Wilson Amplitude showed the best results for both sEMG ($R^2 = 0.952 \pm 0.007$) and iEMG ($R^2 = 0.948 \pm 0.008$), with no significant difference ($P = 0.658$). Application of an advanced model based on artificial neural network did not improve the performance ($P = 0.895$). We have concluded that a linear model is sufficient for force prediction (0-100% MVC), and that iEMG is potentially suitable for proportional control in the same manner as when using a more global measure of intensity.

INTRODUCTION

For many amputees, the only possibility for restoration of movement is through the use of prosthetic devices. Surface EMG (sEMG) is already being used for the control of myoelectric prosthetic devices, where the applied force is estimated proportionally to features extracted from sEMG. [1,2,3] Despite good results, the use of sEMG has a number of disadvantages: (1) it is limited to one or two Degrees of Freedom (DoF), (2) it can only be measured from superficial muscles, (3) it is sensitive to crosstalk, and (4) it can cause irritation of the skin during repeated use.[1,4]

Use of intramuscular EMG (iEMG) for prosthetic devices has been proposed because iEMG may provide more stable and selective recordings compared to sEMG, and may allow effective control of multiple DoFs. Furthermore, iEMG electrodes may be chronically implanted.[1] Because of their high selectivity, iEMG may be less representative of the global muscle activity and thereby contain less information about the force produced by the entire muscle. To our best knowledge, very few features of iEMG have been explored in relation to force e.g. integrated EMG,[5] global discharge rate,[1] root mean absolute values and constraint sample entropy (CSE). [6] Furthermore, no studies have shown whether the used features proposed for sEMG can be applied for iEMG in the entire range of force from 0 to 100% Maximum Voluntary Contraction (MVC).

Therefore, the aim of this study was to assess the predictive capabilities of 14 EMG features for both iEMG and sEMG using the entire force range from 0 to 100 % MVC. This was based on a linear relationship and a relationship found by an Artificial Neural Network (ANN).

METHODS

Experiment

Subjects: The study included 11 right-handed healthy subjects (4 w/7 m) in the age of 22 to 26 years (mean 23.8 yrs). The experiment was approved by the Danish local ethical committee (approval no: N-20080045). All subjects received both written and oral information about the experiment and gave written consent prior to the experiment.

Procedure: The subjects performed power grasping on a force dynamometer (Noraxon) with their right hand, while seated in a chair with their arm placed in an armrest (Figure 1). The MVC force of each subject was recorded three times with a 3 min rest after each trial. The subjects were then asked to follow four different force profiles:

1. A *step* profile of 9 sec with force increasing in 6 steps.
2. A *double ram*p profile of 9 sec.
3. A *bell* profile of 9 sec.
4. A *free varying* profile of 9 sec where 100% MVC had to be reached at least once.

The order of the profiles was randomized. The *step*, *double ramp* and *bell* profile were recorded two times each and the level of force spanned from 0 to 100 % MVC. The *free varying* profile was only recorded once. The force was shown on an oscilloscope in order to provide the subject with visual feedback during each profile. Each trial was followed by a 3

min rest, and all subjects were provided with adequate time to practice matching the profile before the actual recordings.

Data recording

In order to measure grasping force a Jamar compatible handgrip dynamometer (Noraxon) with an adjustable grip size was used.

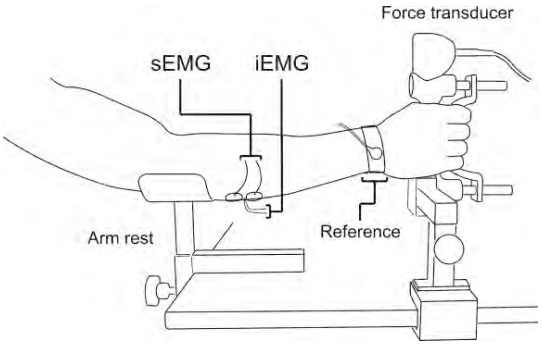

Figure 1: Sketch of the experimental setup.

The grip size for each subject was determined based on which setting resulted in maximum force whilst being comfortable for the subject. The iEMG electrodes (custom-made by use of hypodermic needles and Teflon coated wires (A-M Systems, Carlsberg, WA; diameter 50 μm)) were inserted in the muscle Flexor Digitorum Profundus (FDP) at the middle one-third of the forearm ventral to the ulnar shaft. [7] The electrodes were placed in a bipolar configuration. The analogue output from the iEMG electrodes was amplified with a factor of 1000 and filtered with a bandpass of 20-5000 Hz.

Simultaneously, sEMG (Ambu Neuroline 720) was recorded in a bipolar configuration from the same muscle. The analogue output from the sEMG electrodes was amplified with a factor of 2000 and filtered with a bandpass of 20-500 Hz.

The same amplification and filtering device (EM001-01 SMI) was used for both iEMG and sEMG. A wristband was used as a common reference electrode. The analogue output from force, iEMG and sEMG was sampled by use of a 16 bit AD converter (NI-DAQ USB-6259) with a sampling frequency of 20 kHz.

Signal processing

Digital filters: Apart from the analogue filtering, three digital 4th order Butterworth filters were applied. The force was lowpass filtered with a cutoff frequency of 20 Hz. The iEMG and sEMG were bandpass filtered with frequencies of 100-2500 Hz and 20-500 Hz, respectively. Furthermore, a 2nd order Butterworth filter with a cutoff frequency of 1 Hz was applied to the extracted features.

Extracted features: In total 14 features were chosen to represent the iEMG and sEMG signals. Windows of 200 ms with a step size of 50 ms were applied and features were computed for each window. The same window size was applied to the force signal where the mean was calculated for each window. Thresholds that were general for all subjects and profiles were found by visually inspecting the performance of the features. The extracted features were; *Waveform Length (WL)*, *Zero Crossing (ZC), Slope Sign Changes (SSC), Wilson Amplitude (WAMP)*, *Mean Absolute Value (MAV)*, *Modified Mean Absolute Value (MMAV)*, *Mean Absolute Value Slope (MAVSLP)*, *Variance (VAR)*, *Autoregressive model (AR-model), Histogram EMG (HEMG)*, *EMG envelope energy (EMG_env_energy)*, *EMG envelope (EMG_env)*, *Constraint Sample Entropy (CSE) and Root mean square (RMS)*. See Bøg et al.[8] For further description about implementation of the features.

Data analysis

Force was predicted using two different approaches:

Linear Prediction: For each feature a linear model was derived (with a 1st order polynomial) based on data for all combinations of the *bell*, *step* and *double ramp* profiles. This linear model was then used to predict the force produced during the *free varying* profile.

Artificial Neural Network (ANN): An ANN was used to find the association between each feature and force using data for all combinations of the *bell*, *step* and *double ramp* profile for training. In this study a three layer ANN architecture was applied. The transfer function for the hidden layer was a tan sigmoid and for the output layer a linear transfer function was used.[9] The Levenberg-Marquardt training method was used with the Mean Square Error (MSE) as the performance function. Weights and biases were set randomly at the beginning of the training.[9] The training of the network was done 50 times and the network with the best R^2-value was chosen. The *free varying* profile was then used for testing the model.

Statistical analysis

The statistical analysis was done separately but in the same way for the linear prediction and for the ANN. Moreover the two models were compared. A one-way ANOVA (with factor features) was performed in order to find the feature with the highest mean R^2-value for both sEMG and iEMG. Furthermore, a paired t-test was performed in order to compare the two signals. The comparison of the two models was performed using a paired t-test.

RESULTS

Linear prediction

In Figure 2, the R^2-values for the different features for linear prediction are depicted. *WAMP* showed to have the highest mean R^2-value for both iEMG ($R^2 = 0.948$) and sEMG ($R^2 = 0.952$) with no significant difference between the signals (P = 0.658). For iEMG, *WAMP* was significantly different from *CSE* (P = 0.038) and from *MAVSLP*, *HEMG* and *AR-model* (P < 0.01). For sEMG, *WAMP* was significantly different from *ZC* (P = 0.041) and from *MAVSLP*, *HEMG*, and *AR-model* (P < 0.01).

ANN

The feature with the highest mean R^2-value was *CSE* for iEMG ($R^2 = 0.937$) and *WAMP* for sEMG ($R^2 = 0.927$) with no significant difference between the signals (P = 0.365). For iEMG, *CSE* was significantly different from *VAR* (P = 0.024), and from *MAVLSP*, *HEMG* and *AR-model* (P ≤ 0.001). For sEMG, *WAMP* was significantly different from *ZC* (P = 0.015), and from *MAVSLP*, *HEMG* and *AR-model* (P < 0.001).

Comparing linear prediction and ANN

The paired t-test showed that iEMG had similar mean R^2-values for ANN (*CSE* with *double ramp*, $R^2 = 0.949$) and linear prediction (*WAMP* with *bell-step-double ramp*, $R^2 = 0.948$, P = 0.895). The same result was observed for sEMG.

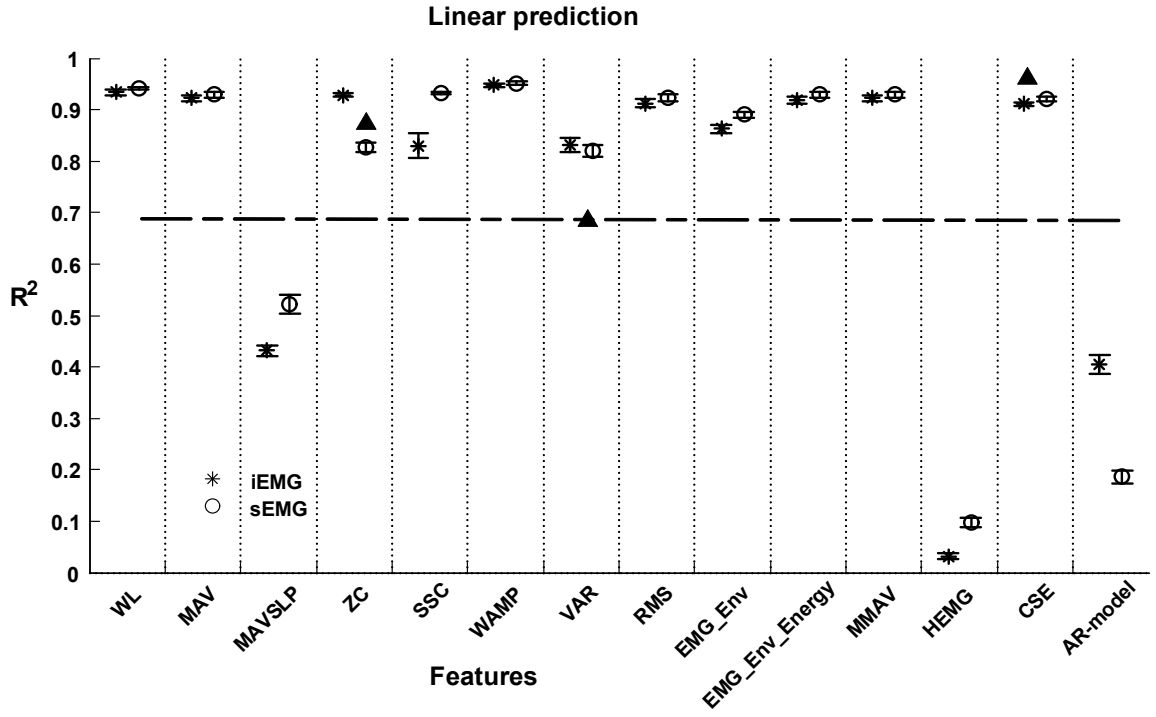

Figure 2: Performance of all features from linear prediction for all profiles for iEMG and sEMG. The x-axis represents the 14 features. The y-axis represents the R2-values with the standard error (SE). The circles and stars represent sEMG and iEMG, respectively. All features below a filled triangle and below the dashed line are significantly worse than WAMP feature.

DISCUSSION

The results showed that it is possible to predict force based on a linear relationship between force and features extracted from either sEMG or iEMG. The relationship and the prediction performance were dependent on the type of feature. Further, results for sEMG and iEMG were similar for both the linear prediction and ANN with $R^2 > 0.9$.

Force prediction

In a study by Phinyomark et al.[10] the *WL* feature showed the best performance for classification of hand movements; however, WAMP also had a good performance. This is similar to the results from the present study, which showed that *WL* had a good performance for force prediction, not significantly different from the best feature for both iEMG and sEMG (*WAMP*). This shows that the WL and WAMP features have an overall good performance and provide a good representation of the muscle activation, regardless of their application. Furthermore, Phinyomark et al.[10] showed that MAVSLP had the worst performance compared to other features, which is also valid for the present study, and therefore the MAVSLP in general provides an insufficient representation of the muscle activation. However, it should be noted, that Phinyomark et al.[10] only evaluated features extracted from sEMG, where the present study investigated both iEMG and sEMG.

In order to clarify whether there exist a better prediction model than the linear, an ANN was used. The ANN prediction showed results similar to linear prediction for both sEMG and iEMG with no significant difference between the two prediction models. The same conclusion was obtained by Kamavuako.[6] Thus, the choice of model (linear prediction or ANN) does not play a significant role when the best feature is selected. However, for our study the performance of the ANN in general seemed to vary, which implies that the possibility of other relationships performing better should be investigated further.

Model selection

Even though the linear prediction in general showed good performance it was not taken into consideration that there might be a difference in the properties of the EMG signals for increasing and decreasing force. Thus, the model was based on only one linear relationship instead of a relationship for increasing force and for decreasing force, respectively. Future work should investigate if there is a difference in the increasing and decreasing EMG-signals, and if necessary, a new model should be defined in order to provide better force predictions.

ACKNOWLEDGEMENTS

This study was supported by a grant from the The Danish Council for Independent Research | Technology and Production Sciences (FTP).

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TOWARDS AN OPTIMAL MODEL FOR THE ESTIMATION OF FORCE FROM INTRAMUSCULAR EMG

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ABSTRACT

In this study we have investigated a potential optimal model for the relationship between muscle force and electromyogram (EMG) that includes both increasing and decreasing force paths (hysteresis).

Intramuscular (iEMG) and surface (sEMG) EMG were recorded concurrently from the muscle flexor digitorum profundus (0-100% MVC) in 11 subjects. Three features, Mean Absolute Value (*MAV*), Wilson Amplitude (*WAMP*) and Constraint Sample Entropy (*CSE*) were computed from the EMG signals. Two models, first (*poly1*) and third (*poly3*) order polynomial, were investigated in two cases: 1) Taking the hysteresis into account for ascending (contraction: *cont*) and descending (relaxation: *relax*) force and 2) Disregarding the hysteresis (*overall)*.

For iEMG the results for *poly1* showed that hysteresis-based models (*cont:* 0.944 ± 0.010, *relax:* 0.939 ± 0.008) had significantly ($P < 0.01$) higher R^2-values (mean ± SE) than the *overall* model (0.889 ± 0.016). For *poly3* a significant difference ($P < 0.01$) was also found between hysteresis-based models (*cont:* 0.963 ± 0.010, *relax:* 0.985 ± 0.002), and the *overall* model (0.926 ± 0.013). Similar results were obtained for sEMG. These results imply the existence of a path dependent model, which may improve the accuracy of force estimation.

INTRODUCTION

Myoelectric prosthetic devices controlled by surface EMG (sEMG) is clinically used to restore some of the lost functions for patients with upper limb amputations. [1,2,3] Intramuscular EMG (iEMG) has been suggested as a potential solution for increasing the number of degrees-of-freedom (DoFs). Apart from the possibility of more DoFs, iEMG also has other advantages such as limited crosstalk and chronic implantation of the electrodes.[1,2] One of usability requirements for a prosthetic device is to make the control as intuitive as possible for the user. This implies for example, providing better proportional control where the level of activation corresponds to the level of muscle activity.[3] Several studies have investigated proportional control where force has been estimated based on features from the EMG signals (See [6] for a review). Common for all these studies is that the proposed model (linear or nonlinear) is computed based on the overall force/feature relationship.[4] Thus, to our best knowledge, none of these studies have investigated if the degree of association between force and features of EMG was dependent on the path of the force profile i.e. one model for increasing force and another model for decreasing force. Deep knowledge of EMG force hysteresis is still missing in the literature.

Ridgway et al.[5] investigated the relationship between force and calcium concentration. It was found that the force was higher when the calcium concentration was decreasing than when increasing, thus hysteresis was found in the force-calcium relationship. Since calcium is needed in muscle contractions, hysteresis might also be present in the feature/force relationship. Therefore, the aim of the present study was to investigate if the feature/force relationship was dependent on the contraction path for both iEMG and sEMG in order to investigate a potential optimal model for future prosthetic devices.

METHOD

Experiment

Data obtained from a previous study [4] was used in this study. Eleven healthy right-handed subjects were included (4 w/7 m, age range 22 - 26 years, mean = 23.8 years). The protocol was approved by the Danish local ethical committee (approval no.: N-20080045). All subjects received both written and oral information about the experiment and gave written consent prior to the experiment.

Procedure: The subjects exerted force while seated in a chair with their right arm placed in an armrest (Fig. 1). First, the subjects exerted MVC force three times with a 3 minutes rest between the trials. Afterwards the subjects were asked to follow a bell-shaped force profile of 9s with force levels ranging from 0 to 100% MVC. Subjects were provided with adequate time to practice matching the profile before the actual recording. The profile was measured twice with a 3 minutes rest in between. [4]

Data recording: In the experiment, a Jamar compatible handgrip dynamometer (Noraxon) with an adjustable grip size was used in order to measure the grasping force. The grip size was set according to the maximum force of each subject. The iEMG electrodes (custom-made by use of hypodermic needles and Teflon coated wires (A-M Systems, Carlsberg, WA; diameter 50 μm)) were placed in a bipolar configuration, in the muscle flexor digitorum profundus (FDP). The needle was placed in the middle one-third of the forearm ventral to the ulnar shaft. The iEMG signals were amplified with a factor of 1000 and filtered with a band pass of 20-5000 Hz. Simultaneously, sEMG was recorded in a bipolar configuration (Ambu Neuroline 720) from the same muscle. The sEMG signals were amplified with a factor of 2000 and filtered with a band pass of 20-500 Hz. A wristband was used as a common reference electrode. Force, iEMG and sEMG signals were sampled by use of a 16 bit AD converter (NI-DAQ USB-6259) with a sampling frequency of 20 kHz.

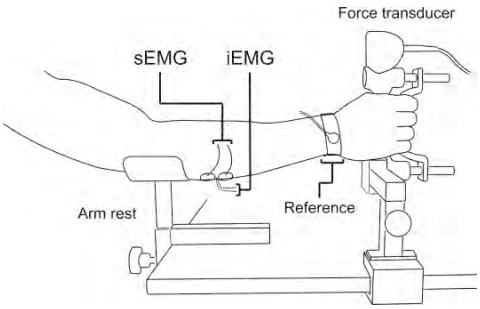

Fig. 1 Sketch of the experimental setup.

Signal processing

Digital filters: A 4th order Butterworth filter was applied for each signal. The force was low pass filtered with a cut-off frequency of 20 Hz. The iEMG and sEMG signals were band pass filtered with frequencies of 100-2500 Hz and 20-500 Hz, respectively.

Extracted features: Three features were chosen to represent the iEMG and sEMG signals; Mean Absolute Value (*MAV*), Wilson Amplitude (*WAMP*) and Constraint Sample Entropy (*CSE*). A moving window of 200 ms was applied to the EMG signals with a step size of 50 ms. Features were calculated for each window. The same moving window was applied on the force signal, where the mean was calculated. Thresholds for computing *WAMP* were found by visually inspecting the performance of the features. The used threshold levels were the same for all subjects and profiles.

Data analysis

The relationship was found between the extracted features of EMG and the corresponding grasping force using two different models for two different cases. The first model was a linear relationship described by a first order polynomial (*poly1*) whereas the second model was a third order polynomial (*poly3*). These models were tested in two cases: Case 1) Taking hysteresis into account for ascending (contraction, *cont*) and descending (relaxation, *relax*) force, and Case 2) Disregarding hysteresis (*overall*). The performance measure used for the relationships was the adjusted coefficient of determination (R^2-value).

Statistical analysis

For each signal type (iEMG and sEMG) and each model (*poly1* and *poly3*) a two 2-way ANOVAs (with factors cases and features) was used in order to compare the performance of the hysteresis (*cont, relax*) with the *overall* model and, moreover, to compare the features with each other. P-values less than 0.05 were considered significant. The Bonferroni–Dunn adjustment was used for multiple comparisons

RESULTS

In figure 2, a representative example of the hysteresis is depicted to show the dependency of the model to contraction path. In table 1 and 2, the results for the different models for iEMG and sEMG are summarized. The hysteresis-based models had R^2-values above 0.94 significantly higher than the *overall* model ($P < 0.01$). Similar results were obtained for sEMG, though *relax* for the hysteresis model was not significantly higher than the *overall* model for *poly1* ($P = 0.07$).

When using *poly1* on iEMG, *WAMP* feature performed significantly better than the *CSE* feature ($P > 0.01$). For *poly1* on sEMG both *WAMP* and *MAV* were significantly better than *CSE* ($P < 0.021$). Furthermore for *poly3* on sEMG, *CSE* was better than *WAMP*. However, when using *poly3* on iEMG, no difference was found between the features.

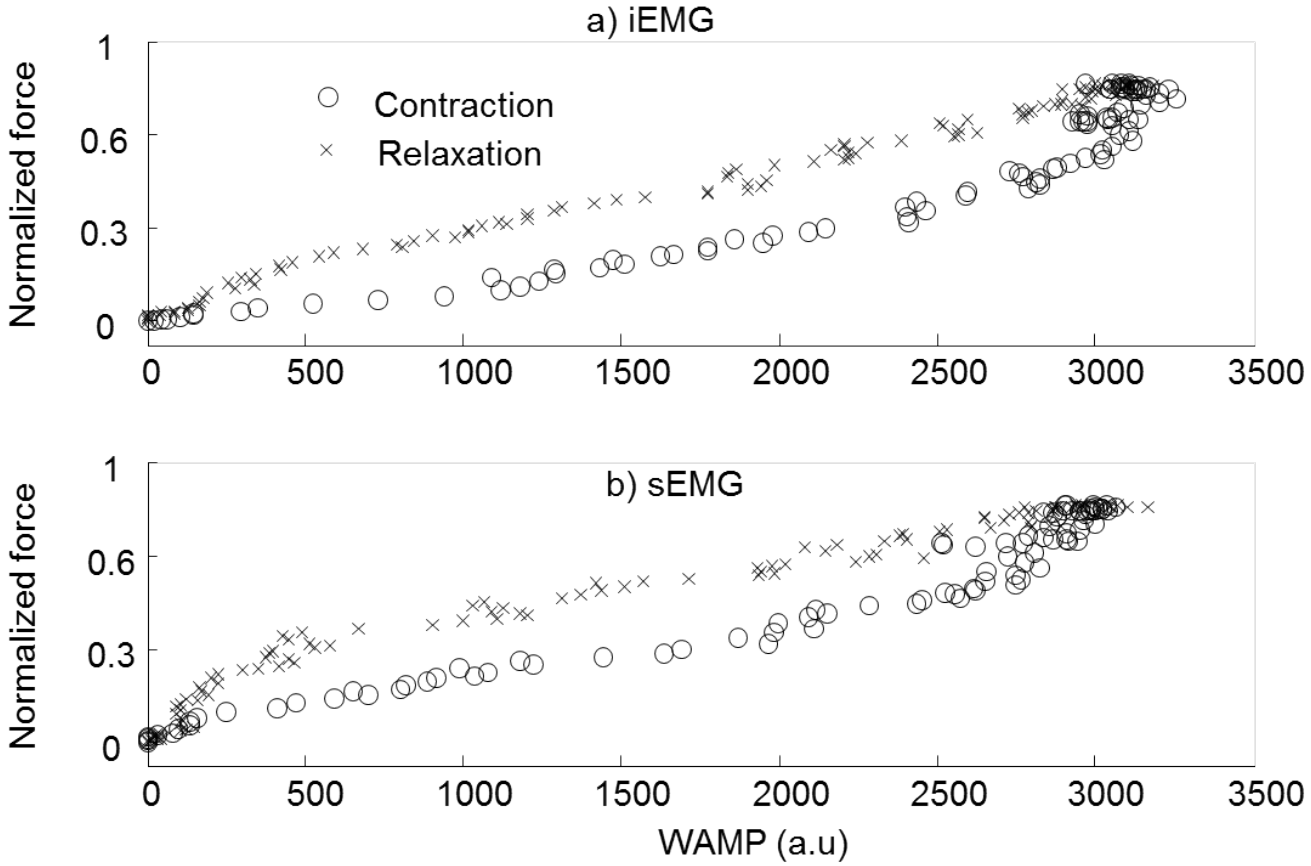

Figure 2: A representative example of the hysteresis obtained using WAMP feature for a) iEMG and b) sEMG. a.u stands for arbitrary unit, circles (o) depicts the ascending force (contraction) path and x *'s the descending (relaxation) path.*

Table 1: Results from the two different models for iEMG when tested in the two cases. The P-values is for the comparison of case 1 (the models for the ascending (cont.) and descending (relax) force) with case 2 (the overall model disregarding hysteresis).

Models for iEMG	R^2	SE	CI	P
Poly1:				
Cont.:	0.944	0.010	[0.922 , 0.966]	0.002
Relax:	0.939	0.008	[0.922 , 0.956]	0.006
Overall	0.889	0.016	[0.853 , 0.925]	
Poly3:				
Cont.:	0.963	0.010	[0.941 , 0.984]	0.008
Relax:	0.985	0.002	[0.980 , 0.989]	0.003
Overall	0.926	0.013	[0.897 , 0.956]	

Table 2: Table 1: Results from the two different models for iEMG when tested in the two cases. The P-values is for the comparison of case 1 (the models for the ascending (cont.) and descending (relax) force) with case 2 (the overall model disregarding hysteresis).

Models for sEMG	R^2	SE	CI	P
Poly1:				
Cont.:	0.949	0.006	[0.936 , 0.962]	0.004
Relax:	0.946	0.006	[0.933 , 0.958]	0.076
Overall	0.925	0.007	[0.910 , 0.940]	
Poly3:				
Cont.:	0.974	0.005	[0.964 , 0.985]	0.006
Relax:	0.985	0.002	[0.936 , 0.962]	0.004
Overall	0.957	0.007	[0.936 , 0.962]	

DISCUSSION

The results showed that the hysteresis-based models were significantly better than the overall relationship, indicating that the relationship between features and force in the full force range is path dependent. As shown in Figure 2, force was higher in the relaxation phase than in the contraction phase with respect to same value of *WAMP* feature. This is to some extend similar to the results obtained by Ridgway et al.[5]. In their study force was higher for the decreasing calcium concentration, indicating a muscle in the relaxation path.

This study only focused on comparing the two hysteresis based models, *relax* and *cont,* with an *overall* model and therefore the computed models, *poly1* and *poly3*, were not compared. Moreover, the same type of relationship was used both for the increasing and decreasing force. Other types of models might show even higher R^2-values and probably the two paths have different relationships, which should be investigated further. The search for an optimal model is further emphasized by the fact that the performance of features is model dependent. Not all features perform equally using *poly1*, thus for every feature used the optimal model should be investigated to maximize the association with force.

In the present study, we only found the relationship between EMG features and force, and did not predict force based on these relationships. Thus, even though *poly3* gave the highest R^2-values, it may be over-fitting the data and might therefore perform less effective when used for prediction.

In conclusion, this study showed strong indications (for all subjects) of hysteresis in the relationship between EMG features and force which is a step towards an optimal model for force estimation.

ACKNOWLEDGEMENTS

This study was supported by a grant from and the Danish Agency for Science, Technology and Innovation (Council for Independent Research | Technology and Production Sciences, Grant number 10-080813)

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RESOLVING THE LIMB POSITION EFFECT

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INTRODUCTION

Electromyography (EMG) has been used as a control input for powered upper limb prostheses for decades. Alternative biosensors, like myokinemetric sensors [1],[2], mechanomyographic sensors [3] and accelerometers [4] have been used for upper limb pattern recognition in more general terms but have not produced accuracies acceptable for prosthetic use.

The desire to use a larger number of myoelectrode sites to facilitate control of multiple degrees of freedom has been counteracted by the added complexity, cost, space, and weight associated with additional sites. Thus, commercial upper limb prostheses today usually have only two electrode sites, while researchers continue to experiment with multiple sites [5]. An alternative to the uni-modal EMG approach for increasing the degrees of freedom is a multi-modal approach. Instead of adding additional EMG channels, it is possible to combine EMG and other sensor modalities (e.g., force sensors [6] or accelerometers [7]) in order to improve pattern recognition performance. Other examples of multi-modal solutions exist [8], [9].

In our previous work [10] it was shown that variations in limb position associated with normal use can have a substantial impact on the robustness of myoelectric pattern recognition. We proposed to solve this problem, hereafter referred to as the *limb position effect,* by training the classifier in multiple positions and by measuring the limb position with accelerometers. Applying these methods to data from normally limbed subjects, the classification errors were reduced substantially.

In the present study, we have examined the generalizability of the training set as a function of the number of training positions in the set. This makes it possible to define a minimum training procedure, in order to reduce the training time for the end user.

Finally, we have investigated accelerometers as a supplementary modality for EMG. Accelerometers are relatively cheap, small, robust to noise and easy to integrate in a prosthetic socket. This work examines the efficacy of accelerometers in comparison to adding expensive and space-consuming electrode sites.

METHODS

All experiments were approved by the University of New Brunswick's Research Ethics Board.

Population and Data Acquisition

EMG data corresponding to eight classes of motion were collected from 17 healthy normally limbed subjects (10 male, 7 female) within the age range 18 to 34 years.

Subjects were fitted with a cuff made of thermo formable gel (taken from a 6mm Alpha liner by Ohio Willow Wood) that was embedded with eight equally spaced pairs of stainless steel dome electrodes (EL12 by Liberating Technologies, Inc.). The cuff was placed around the dominant forearm (13 right, 4 left), proximal to the elbow, at the position with largest muscle bulk. A reference electrode (RedDot by 3M) was placed over the back of the hand. Two analog 3-axis accelerometers (Freescale MMA7260QT MEMS) were used to estimate limb position. The first accelerometer was affixed adjacent to the cuff on the forearm, over the brachioradialis muscle. The second was placed over the biceps brachii, aligned with the forearm accelerometer when the subject was reaching forward (see position P2 in Fig. 1). Both accelerometers were configured to have a sensitivity of 800 mV/g at a range of ±1.5 g, where g represents acceleration due to gravity.

The eight channels of EMG were differentially amplified using remote AC electrode-amplifiers (BE328 by Liberating Technologies, Inc.), and low pass filtered at 500Hz with a 5th order Butterworth filter. Finally, the six accelerometer channels and eight EMG channels were acquired using a 16-bit analog-to-digital converter (USB1616FS by Measurement Computing) sampling at 1 kHz.

Subjects were prompted to elicit contractions corresponding to the eight classes of motion shown in Table 1**Error! Reference source not found.**. Performance was evaluated using all eight classes, as well as a reduced set of

five classes. This five class system only included classes C3, C4, C5, C6, and C8, which are representative of contemporary powered prostheses. The five class system is referred to as the *contemporary* system and the eight class system as the *advanced* system.

Table 1: Motion classes

C1.	**Wrist flexion**	**C5.**	**Open hand**
C2.	**Wrist extension**	**C6.**	**Power grip**
C3.	**Pronation**	**C7.**	**Pinch grip**
C4.	**Supination**	**C8.**	**Hand at rest**

Each contraction was sustained for three seconds and a three second rest was given between subsequent contractions. Ten trials were recorded in each of the following limb positions (P1–P5; as illustrated in Fig. 1), resulting in a total data set of [n subjects × 10 trials × 5 positions × 8 classes × 3 seconds], where n is explained in Section C.

Fig. 1: Limb positions.

Subjects were instructed to perform contractions at a moderate and repeatable force level and given rest periods between trials to avoid fatigue. The average duration of the experiment (with 50 trials lasting 48 seconds each) was approximately 80 minutes per subject. Some patients noted minor shoulder (deltoid) fatigue.

Data processing

As this work represents an introductory examination of multi-modal pattern recognition, it was appropriate to test the effects using a known control scheme. Englehart and Hudgins [11] showed that simple time-domain (TD) feature extraction combined with a linear discriminant analysis (LDA) classifier can be used as an effective real-time control scheme for myoelectric control. Because of its relative ease of implementation and high performance, this system has been widely accepted and was therefore adopted in the present study. EMG data were digitally notch filtered at 60 Hz using a 3rd order Butterworth filter in order to attenuate any power line interference. Data were segmented for feature extraction using 250 ms windows, with processing increments of 50 ms. The TD features (mean absolute value, zero crossings, number of turns and waveform length) were extracted from the EMG data. Please refer to [11] for details of the feature extraction and the classification.

For each processing window, the average value of the accelerometer data was calculated. Where applicable, this feature (hereafter called ACCEL) was input to the LDA classifier separately or as an extension of the original feature set.

Data exclusion

Some of the subjects were not able to perform consistently throughout the data set. Similar phenomena occur in real-life situations where some individuals have great difficulty producing distinct EMG signals [12]. To ensure consistent data, subjects whose intra-position classification error exceeded 10% (five of the 17 subjects) were excluded from the study. This does not detract from the focus of this work; to ascertain the effects of position on performance. It simply eliminates possible confounding factors that may have been present with those subjects that did not perform well.

In two of the remaining 12 subjects, hardware problems caused erroneous accelerometer readings. Thus, 10 subjects were used in this study.

Classification

The following classifier training schemes were explored:

1) Training in a single limb position

   TD features recorded from a single limb position were used to train the classifier. The classifiers were trained using data from the first five trials and tested using data from the last five trials.

2) Training in multiple limb positions

   TD features recorded in multiple limb positions were concatenated and used to train the classifier. The classifiers were trained using a data set of reduced size per position, so that the total training set size was the same as in 1), in order to make the results comparable.

3) Training with TD and ACCEL features

   TD and ACCEL features recorded in multiple positions were concatenated and used for motion classification. The data set was reduced in the same way as in 2) in order to make the results comparable.

"Leave-One-Out" training strategy

In order to investigate the generalizability of the training set as a function of the number of training positions in the set, the following procedure was employed. For each test position, all possible subsets of the remaining positions were applied as a training set.

Input selection

A signal feature selection scheme was chosen in order to examine which electrode sites and accelerometer signals would be most useful for the pattern recognition. Starting with just one sensor, the best one was chosen (based on the

classification error averaged over all subjects and motion classes). It was then tested in combination with each of the remaining sensors, and the best combination was chosen before adding the next sensor. In this manner the sensors were added to the system one by one.

RESULTS

Training in a single limb position

Five different position-specific classifiers were trained; each one using data from only one of the limb positions, but tested using data from all positions. The resulting intra-position and inter-position errors are shown in Table 2.

Table 2: Intra- and inter-position classification errors for the advanced system, trained in a single limb position, and averaged across all subjects and classe

Intra-position classification error	**3.8%**
Inter-position classification error	**21.1%**
Overall classification error	**17.6%**

Training in multiple positions

In Fig. 2 we present a comparison of how training in multiple positions affects the classification, for the *advanced* system. We have used the *Leave-One-Out* strategy as described in the *Methods* section, part E, in order to investigate the generalizability of the training set as a function of the number of training positions in the set.

Notice that the classification error improvement when increasing the number of training positions from one to two is larger than when increasing to three or four training positions.

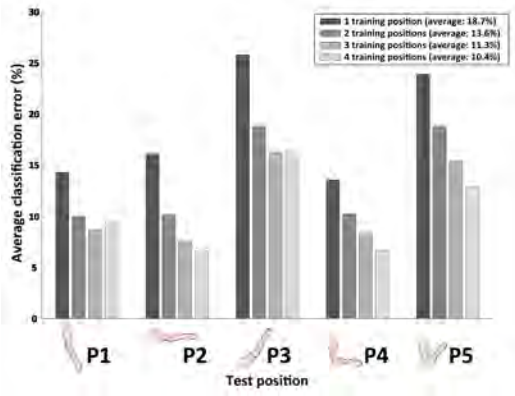

Fig. 2: Comparison of classification errors when testing in one limb position and training in all possible subsets of the remaining positions (the "Leave-one-out" strategy, as described in the Methods section, part E). Note that the training sets have been scaled so that they have identical size every time; independently of the number of training positions, by using subsets of the ten trials.

Relative importance of position information and surface EMG

The results of the input selection described in the *Methods* section, part F, are presented in Fig. 3. It is noteworthy that when adding new sensors one by one, the forearm accelerometer provides more novel classification information than even a second or third EMG electrode. It is also worth noting that the upper arm accelerometer is one of the least useful sensors. This is a desirable result as it would be difficult to justify including a sensor external to the forearm socket, and across the elbow joint.

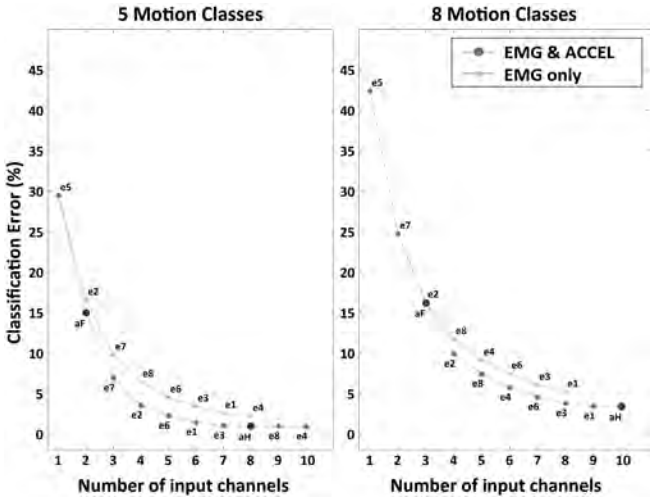

Fig. 3: Classification error as a function of selected input channels, for pattern recognition systems with 5 and 8 motion classes, choosing input channels among 8 electrode pairs (e1–e8) and 2 accelerometers (aF–Forearm, aH–Humerus).

For the *contemporary* system, the improvement flattens out after 4-5 electrodes and one forearm accelerometer (reaching an average accuracy of 98-99%). The advanced system can exploit 6-7 electrodes and one forearm accelerometer (reaching an average accuracy of 95-96%).

DISCUSSION

EMG TD features and training an LDA classifier in in a single limb position yielded an average intra-position error (3.8%) significantly lower than the corresponding inter-position errors (21.1%). These results indicate that EMG classification error is strongly dependent on limb position.

We have shown that the limb position effect can be partially solved by training the classifier in multiple positions. Since training in multiple positions can be cumbersome for the end user, it is however desirable to reduce the number of training positions. Therefore it is an advantage that most of the improvement is achieved already when increasing from

one to two training positions (reducing the average error from 18.7% to 13.6%). Previously we have also shown [10] that it is important to have a training set containing a variation of elbow angle.

The accelerometer lends itself to being used in human-machine interfaces due to its small size, low cost, and simple mechanical and electrical interfaces. The absence of many of the disturbances often encountered in EMG sensors and similar devices makes it interesting as a supplementary sensor in hand motion classification systems, including upper limb prostheses.

The accelerometer does not provide an estimate of muscle force, but we have shown that it provides useful information that can supplement EMG signals. If one wants to improve a system originally having two EMG electrodes, a multi-modal approach can be taken. The results demonstrate that it is more advantageous to add an accelerometer affixed to the forearm (multi-modal approach) rather than increase the number of EMG channels (uni-modal approach).

Even though the limb position effect was discovered and observed in users in the clinic [7], [10], and was resolved for the normally limbed subjects in our study, it needs to be examined specifically for the end users. Gravitational and biomechanical effects of limb position will be different for prosthetic users compared to the normally limbed subjects of this study. As such, we are planning to extend this study to include prosthesis users.

This work is part of a larger investigation aimed at improving the practical robustness of myoelectric control. The present results indicate that facilitating position invariant myoelectric control through methods such as feature selection, data projection, multi-sensor systems, or by other means could be an important part of this larger work.

ACKNOWLEDGEMENTS

The authors would like to thank P. J. Kyberd for important input concerning the experimental protocol and invaluable practical assistance in conjunction with the accelerometers.

The work was supported in part by NSERC Discovery Grant 217354-10, NIH Grant 5R01HD058000-03, the Atlantic Innovation Fund, Strategic Area Medical Technology at NTNU and The Research Council of Norway (Grant 192546).

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ELECTRICALLY CONDUCTIVE SILICONE INTERFACE FOR MYOELECTRIC PROSTHESES WITH SILICONE SOCKET SUSPENSION

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ABSTRACT

Silicone socket suspension technology for the upper extremity amputee has been proven to provide increased suspension and range of motion over conventional self suspending sockets. Application of silicone socket suspension has greatly benefitted clients with very short residual limbs and disarticulation level amputations. However, the use of silicone suspension sockets with myoelectric control has presented some problems, namely relating to the wear and tear on electrical control cables and connections. Managing the interface between silicone socket and hard socket without compromise to signal loss via an electrical-mechanical interface about the pin lock has been investigated, and although function was acceptable, this approach was technically complex. An alternate and potentially simple solution is the use of an electrically conductive silicone interface within the silicone socket and localized at the electrode site. To investigate the feasibility of this approach, a prosthetic socket and silicon sleeve using the conductive material was fabricated and evaluated on a single subject. Signal quality was found to be acceptable but further work is needed to assess the factors that can be targeted to further improve the signal-to-noise ratio. This approach has the potential to reduce the technical requirements in achieving usable EMG signal capture.

CODING SCHEME FOR CHARACTERISING GAZE BEHAVIOUR OF PROSTHETIC USE

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INTRODUCTION

During the performance of a task in which detailed visual information is central to its success the centre of the visual field (located at the fovea) is continuously focused on key visual cues within the scene [1]. Studies suggest that the visual behaviour during the performance of a novel and challenging upper limb task changes as subjects become skilled1 [3]. For example, during the early stages of learning to use a hand-operated tool, gaze closely monitors the tool movement towards the target, to obtain visual feedback on its location. With training, the relationships between arm movements and tool location are established and thus tool location can be predicted from proprioceptive feedback. This leads to a change in gaze behaviour, with fixation increasingly moving to the target, rather than following the tool.

The changes to gaze behaviour over the course of skill acquisition have a number of potential clinical applications. For example, comparing gaze behaviour of trainees with gaze behaviour of experts may provide useful insight into trainees' performance [4]. Gaze might also be used as a training tool in itself. For example, a study of novice basketball players showed that improvements in performance resulted from observing gaze behaviours of expert players [5].

In studies of gaze behaviour, the location of the foveal focus is typically estimated from data collected from a head-mounted camera monitoring the eye. These data are used to project a symbol (typically a cross-hair) onto a scene video, which is collected from a second head-mounted camera. In order to interpret gaze data a method is required for describing and summarizing the trajectory of the foveal gaze within the scene video.

In previous work on gaze behaviour in the performance of everyday tasks (i.e. in unstructured environments), gaze has been described in terms of periods spent focusing on Areas of Interest (AOI). Areas of Interests (AOIs) in the scene video typically consisted of a set of objects that the eye was focused on during the task performance. Although more recent work has begun to consider the functional implications of focusing on different parts of objects, there are very few examples in the literature of clearly described coding schemes that allow for such behaviour to be unambiguously described. Describing gaze data without a predefined coding scheme is likely to make the process open to personal interpretation of the rater.

The gaze location is projected on to a 2D video of the scene containing no information about the depth. This discrepancy in the dimensionality might make also the gaze location open to misinterpretation and hence poor inter-rater reliability. For instance, if two AOIs overlap with each other in the line of sight, a common occurrence in manual tasks, the cursor would be projected on to the object that is closer to the subject. However, there is ambiguity in some cases. For example, in cases where gaze is focused on one object and, a second object is moved to partly obscure vision of the first object, it is difficult to judge whether the subject is taking information from the near, or far object.

In a related paper [6], we report on a study of gaze behaviour during the performance of a functional task (pouring water from a carton to a cup) in anatomically intact subjects learning to control a myoelectric prosthesis. In this paper we describe the development and validation of an objective gaze coding scheme for characterization of gaze behaviour during performance of the carton pouring task By describing the process by which the scheme was developed we provide the potential to generalise the approach to other similar tasks. AOIs in the scene are strictly defined, based on a functional interpretation. A method is proposed for dealing with uncertainty in AOIs arising from the dimensionality discrepancy between the 3D scene and 2D gaze video data. Finally, we report an inter-rater reliability study demonstrating the reliability of the proposed coding scheme.

METHODS

Subjects

Following ethical approval from the University of Salford's Research Ethics Committee, 2 right-handed anatomically intact male subjects (28 and 30 years) who had normal-to-corrected acuity and colour vision were recruited

[1] We define skilled motor behaviour as the ability to predict the consequences of physical actions [2]

for this study. Prior to admission to the study, all subjects signed an informed consent form.

Manual task performance and gaze tracking

Gaze data were gathered using the iView X™ HED 2 (SenseMotoric Instruments GmbH, Tellow, Germany) Eye-Tracking system.

The subjects sat with their back straight, supported by the back rest of a chair, with both upper arms abducted by approximately 30º, elbows in about 90º flexion and with hands resting comfortably on top of the table (Figure 1). The subject was asked to complete a well-defined, everyday functional task "pouring liquid from a carton into a glass" using their left hand (non-dominant hand)2. The carton was placed at a location that could be comfortably reached by the subject, without leaning forward. This location was marked for use in subsequent trials.

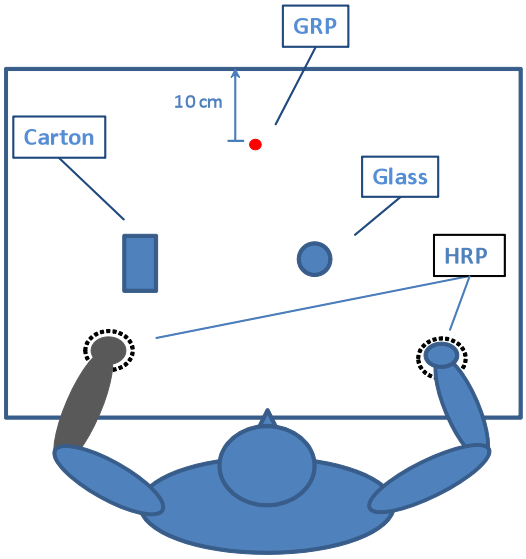

Figure 1: Experimental setup

Subjects were instructed to initiate the task from the hand reference points (HRP, Figure 1) and to return to the reference points at the end. The subjects first reached for the carton, then grasped it, transported it towards the glass, poured a fixed volume of water from the carton into the glass, returned it to its original location, then released the carton and returned the hands to their original positions. This task requires the subject to pay visual attention in order to not spill the water. Further, the carton was easily deformable, potentially adding to the task difficulty.

Subjects were instructed to gaze at a marked point (termed the gaze reference point or GRP) prior to initiating the task and at the end of the task (Figure 1). During task completion, subjects were free to move their eyes as they wished. Furthermore, no constraint on head movement was applied during the task performance.

Data collection was completed over two separate testing sessions approximately 3 days apart; in the first session the task was performed using their left arm; in the second session subjects used a myoelectric prosthesis, fitted over their left arm (Figure 2) to complete the task. The myoelectric prosthesis was equipped with a single degree of freedom electrical hand (RSLSteeper "Select" Myo Electric hand (size 81/4)), whose opening and closing was controlled via EMG signals from a socket-located electrode (for more detail see [7]). In each session, subjects completed the manual task as described above 10 times. Subjects were instructed to perform the task at their own pace. Prior to the second session, the table was moved away from the chair by a suitable distance to accommodate the extra-length of the prosthesis.

Development of the coding scheme

In gaze analysis the scene ahead is typically subdivided into discrete areas of interest (AOIs). Most of the researchers who have studied the performance of functional tasks have defined the set of AOIs as being the set of objects in the scene (see Land et al [1, 8], Hayhoe [9]). Preliminary analysis of our data from the first session showed similar trends to those reported by Land, Hayhoe and others ([1, 8], Hayhoe [9]) with a characteristic sequential pattern of fixation on objects, which could be assumed to contain the necessary visual cue to perform the task [10]. However, close inspection of the gaze data shows that the fixation occurred at specific areas on the objects that appear to have particular functional importance. This fixation on specific areas of objects was also observed in a study by Johansson et al [11] and in our earlier pilot work [10]. Thus, certain areas on the objects seem to be of more importance to the completion of the task than the rest of the object. This suggests that coding schemes that consider only the focus on a particular object as a single unit may be losing useful information.

Therefore it was decided to divide the area that an object occupies into a number of AOIs. The set of AOIs was determined following a series of discussions following pilot data collection, based on the assumption that AOIs should have functional relevance. In addition, three further AOI's were defined that were not part of an object, but functionally related to the nearby object: eye "Following the hand", eye "Following the carton" and Above Carton. These AOIs

A total of 14 AOIs were identified, as well as a "missing data" category (to account for saccades, blinks and periods when gaze location was undefined), as shown in Figure 2.

[2] The study was limited by the availability of only left-handed myoelectric hands within the Department

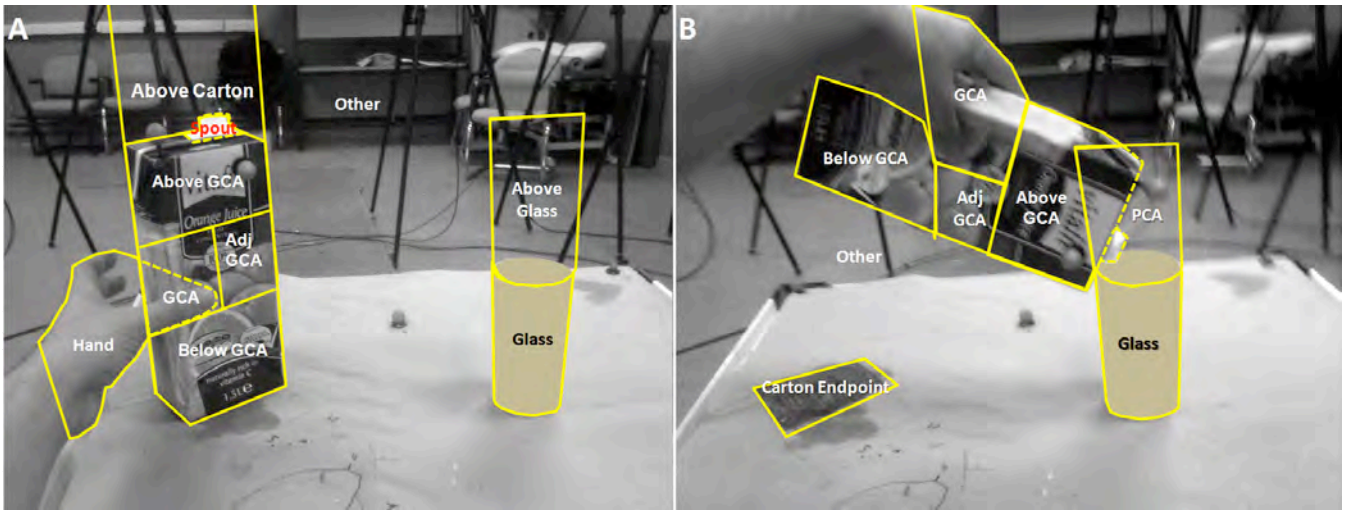

Figure 2: The areas of interest (AOIs). Note – "Following Hand" and "Following Carton" are not shown.

CODING SCHEME INTER-RATER RELIABILITY

Following the development of the coding scheme, the data from the two subjects were coded using BeGaze software (SensoMotoric Instruments GmbH, Tellow, Germany) that comprises a built-in algorithm to discriminate fixation periods from other periods (saccades and blinks). Two raters (M and R) were invited to separately code gaze data for both subjects. Each rater was asked to firstly define the onset and the end of the task based on the hand movement and then to record the temporal sequence of gaze on AOIs, as well as the time spent on each AOI. The mean task duration is listed in Table 1 for the two raters. A t-test showed no significant difference in task duration between raters (*p-value* = 0.684).

Table 1: Mean (SD) of task completion duration in seconds as measured by the two raters.

Subject (condition)	Rater M	Rater S
Subject 1 (anatomical hand)	10.1 (0.9)	10.3 (1)
Subject 1 (Prosthesis)	18.2 (2.5)	19.1 (3.4)
Subject 2 (anatomical hand)	10.5 (1.1)	10.4 (0.9)
Subject 2 (prosthesis)	18.9 (3.6)	18.6 (3.8)

Figure 2 shows examples of the gaze sequence and normalized fixation duration of subjects 1 and 2 using the anatomical hand, and the prosthesis, as coded by each rater.

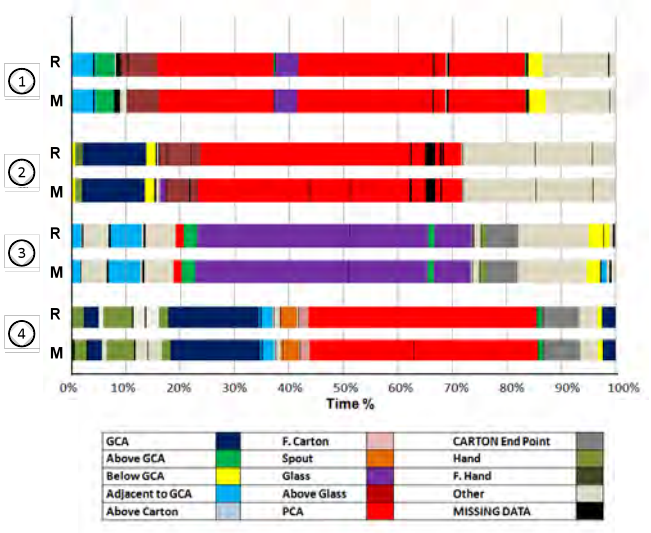

Figure 2: Examples of gaze sequence as coded by raters M and R: (1) subject 1 anatomical hand, (2) subject 1 prosthesis, (3) subject 2 anatomical hand, (4) subject 2 prosthesis.

Table 2 gives the total gaze duration at each AOI (the sum of total fixation duration) and frequency of fixation at each AOI for each rater for all trials.

The interclass correlation coefficient (ICC) was used to compare the total fixation duration at each AOI for each coded trial between the two raters. The 2-way random absolute agreement between the raters was highly comparable as revealed by the ICC (ICC = 0.975, p-value <0.001), with high internal consistency (Cronbach's alpha coefficient = 0.987).

Table 2: Total gaze duration and frequency of gaze fixation at each AOI.

AOI \ Rater	Total Gaze Duration [s]		Frequency	
	M	R	M	R
Grasping Critical Area (GCA)	37.2	38.1	45	50
Above GCA	21.9	21.7	66	68
Below GCA	5.1	5.3	31	29
Adj GCA	3.6	3.8	17	18
F Carton	0.3	0.3	1	1
Spout	7.2	7.2	37	36
Glass	21.2	18.2	43	43
Above Glass	8.1	10.9	56	59
Pouring Critical Area	101.9	106.6	36	37
Carton Endpoint	7.1	8.8	23	19
Hand	12.7	11.2	24	18
F Hand	1.9	1.9	54	48
Above Carton	0	0	0	0
Other	51.0	47.2	154	138
Missing Data	15.7	16.2	253	262

DISCUSSION

Eye tracking offers an object method to explore visual attention. However, gaze coding is usually carried out by visual inspection of the data, sample by sample, and judging which AOI is being hit. The coding scheme therefore was developed to address the subjectivity that the coding process entails.

Interestingly, the distribution of focus on such areas appears to change when the prosthesis arm is introduced [10]. Therefore, a coding scheme that simply considered objects as the AOIs, would fail to account for the observed changes in fixation patterns.

Nevertheless and despite the effort to eliminate the subjectivity of gaze coding, in a few cases location of the gaze fixation was observed to still depend on the rater's opinion. For instance, when the gaze is fixating marginally between adjacent AOIs, the rater has to decide which AOI to consider. Furthermore, the confusion matrix might not be fully optimized for this task. However any misinterpretation should be consistent and hence can be seen as a source of systematic error.

As the main interest of the reliability investigation is to explore the agreement of the raters to code gaze under the testing conditions, gaze data of both subjects were treated as one sample of independent variables that was rated by two independent raters.

These results demonstrate the reliability of the coding scheme.

CONCLUSIONS

A coding scheme with function-related AOIs has been developed. This study reports, to our knowledge, the first attempt to produce a detailed and reliable coding scheme that incorporate sub-parts of objects as AOIs defined by function, and this is likely to be of interest to researchers studying gaze during complex motor activities. Although defined AOIs might be exclusively applicable to our task, the method used to define AOIs in this work can be generalized.

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A NEW VOLUNTARY CLOSING HOOK PROSTHESIS

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INTRODUCTION

A person with an arm defect can choose between an electrically powered and a body-powered prosthesis. Both options have their own inherent advantages and disadvantages. The electric prosthesis can be esthetical and easy to use, but is heavy, expensive and vulnerable. The body-powered prosthesis is cheap and reliable, but requires an uncomfortable shoulder harness to be operated.

Over a decade ago, a prototype of a new voluntary closing (VC) prosthesis was conceived [1]. It does not need external power or a shoulder harness to be operated, combining advantages of both types of prostheses. This is realized by using passive flexion of the prosthetic wrist to power the prosthesis. An integrated locking mechanism allows the user to hold an object without exerting any operating force. The operating principle is illustrated in Figure 1.

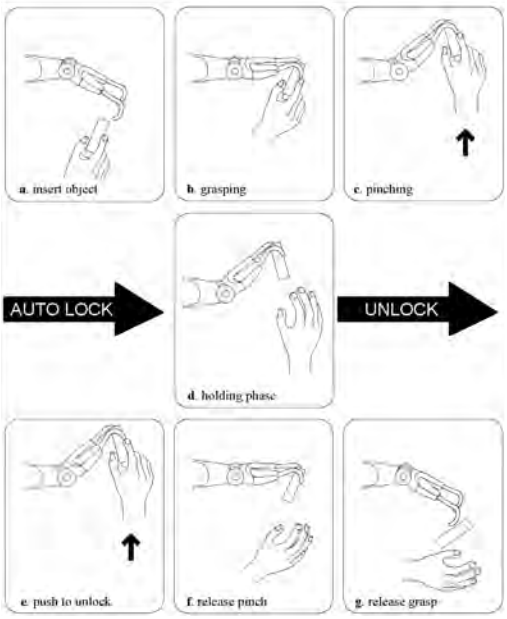

Figure 1: Working principle of the new VC prosthesis, showing the grasping (a, b, and c), locking (d) and releasing/opening (e, f, and g) of the prosthesis.

The new VC prosthesis works excellent for indirect prehension and direct prehension of big or fixed objects. However, direct prehension of smaller objects is very hard to achieve, because there is nothing to push the wrist into dorsal flexion. This does not seem to be a real disadvantage in daily life, as research shows that a very large percentage of activities of daily life is done single-handed [2 – 4]. Almost all two-handed tasks are of such a nature that one hand is holding the object and the other hand is manipulating the object [2, 3]. For these situations a prosthesis only capable of holding an object would suffice. Direct grasping and manipulating with a prosthesis are very rare.

Overall, the new VC prosthesis concept seems very promising. The concept provides a combination of the advantages of both electric and body powered prostheses. This means a lightweight, cheap and reliable prosthesis with an unlimited power source, but no shoulder harness and the ability to hold an object without exhausting the user. The theoretical disadvantages are relatively unimportant in daily life and the advantages are very pronounced. Unfortunately the current prototype is merely showing the concept and is not usable in daily life. This paper briefly describes the design steps taken to transform the concept into a working prototype.

DESIGN CRITERIA

Prostheses must fulfill many criteria to be functional. Obviously there are the criteria that are viable for both mechanical as well as electrical prostheses. These include criteria regarding weight, esthetics, costs, etc.

The new VC prototype will be built in a small size, approximately for children of the age of 4. This means that all dimensions based on adults should be roughly halved [5].

Grasping force

The subject of grasping force is not widely studied in the prosthetic literature. Not only because of the small target group, but also because the needed grasping force heavily depends on the shape and material of the prosthetic device, and the shape and material of the object to be grasped.

Therefore, it is hard to make generalized statements about the needed grasping force.

In a study focused on 2 to 4 year old children [6], measurements on existing prostheses and able bodied children were compared to find both the available grip force from prostheses and from able-bodied subjects. The results indicate a grasping force of 26N to 69N for the able-bodied children. Another study [7] states a maximum pinching force of 100N to be typical for commonly used adult sized electrically powered prosthetic hands. Scaling down to the child size results in a minimum grasping force of 25 Newton for four year old able bodied children.

Another report [8] concludes that 2-year-olds can hold most objects if they have at least 9 N of grip force; 3 and 4-year-olds require 18 N of grip force for their activities.

Grip opening

The needed grip opening depends on the objects gripped in daily life. Relevant sizes of objects were measured to determine the needed grip size. The results suggest a maximum needed opening width of 70 to 80 mm for adults. For children 4 years of age this results in a maximum needed opening of 35 to 40 mm, which is in the same order of magnitude as reported in [2].

Operating force

The force that can be exerted on the prosthesis by the other hand when grasping indirectly, or on the surroundings when grasping directly, determines the operating force. Unfortunately, no data are available about the forces that subjects can produce in this manner. Therefore, measurements were taken from adult, able-bodied subjects. The force one can exert on a table surface in a sitting position was measured, as well as the force one can exert between the two hands. These represent respectively the most common situations of direct and indirect grasping. All measurements were done while the subjects were sitting in front of a table. Every time, the maximum force has been measured. The measurements indicate an average maximum force of 130 N against a table surface, and a maximum force of 200 N between the two hands. According to Monod [9] 18% of this maximum force is an acceptable level for prolonged use. This would result in a comfortable operating force of respectively 23 and 36 Newton. Scaling back to children of the age of four this means a maximum operating force of respectively 32 and 50 Newton and a comfortable operating force of respectively 6 and 8 Newton.

Operating stroke

The operating stroke is the amount of travel used by the mechanism to be operated. This is the length of the dorsal flexion arc of the prosthesis. Naturally, the operating stroke should be as small as possible. A long stroke will result in an awkward angle of the prosthesis when the object is grasped, especially when the object to be grasped is small. The goal is a dorsal flexion that is both visually and functionally as natural as possible.

The operating stroke will be determined by the required grip opening and the transmission ratio needed to produce a sufficient grasping force with the given operating force. Therefore the operating stroke will be a result of other design choices and criteria.

Size and weight criteria

Both arms need to be of the same length to accommodate natural use. Therefore, the length of the sound hand dictates the length of the prosthesis. The length of the mechanism proximal of the wrist should be as short as possible, as this will limit the use for someone with a long arm remnant. The desired goal is to have nothing protruding proximal of the wrist rotation unit.

Considering the symmetry of the amputees' body, the weight of the prosthesis should be as close as possible to the natural weight of a human hand. In the case of a four year old, this is approximately 50 grams, based on volume measurements. In real life the connection between the prosthesis and the amputee is far from ideal and therefore as light as possible is preferred above a realistic weight. For a child's prosthesis this is a very ambitious goal. Therefore, this is not considered a solid demand for a good design, but merely a desired goal.

Practical considerations

The mechanism should be able to withstand the contamination and abuse of daily use, i.e. perspiration, rain, dirt/sand. This is very important, as one of the most critical factors in device abandonment is a lack of reliability [10].

The prosthesis should be able to withstand the loads applied in daily life. In case of a 4 year old, this also includes suspending the body weight from the prosthesis. Therefore, 200 N is considered the maximum load.

To operate the prosthesis, it should not be necessary to use any extra controls, as a need for a second hand would would defeat the purpose of a prosthesis for two-handed tasks. Therefore the locking and unlocking of the grasping has to be operated automatically when applying the operating force.

DESIGN SOLUTION

For the design a hydraulic solution was adopted, incorporating a mechanism capable of switching mechanical

advantage [11]. In the first phase, the connection between both parts is direct, but in the second phase the system switches the flow through a pressure intensifier. The switching moment is determined by the pressure in the system. Locking can be done by closing the feeding line between the master cylinder and the grasping cylinder. A scheme of the complete system is shown in Figure 2.

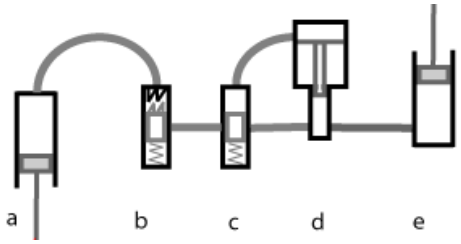

Figure 2, Hydraulic concept with master actuator (a), locking mechanism (b), pressure switch (c), pressure intensifier (d), and slave actuator (e)

Due to the nature of hydraulic systems, the orientation of the separate components is more or less free to choose, allowing a small overall size.

Locking mechanism

A key feature of the new VC prosthesis is the possibility of locking the grasp. It is desired that the locking activates automatically when an operating force is applied. The mechanism used strongly resembles the internals of a ballpoint pen, Figure 2b. Pressure on the master cylinder will force the internals of the locking mechanism to move upwards. Releasing the pressure on the master cylinder will force the internals of the locking mechanism back. Due to the interlocking teeth, the internals will rotate with each time the master actuator is operated and released. After each rotation of 90 degrees, or one push and release of the master actuator, the system will go from unlocked to locked, or vice versa.

Two-phase system

The basic principle of hydraulic two-phase mechanisms, or pressure intensifiers, is pressure enhancement by connecting two pistons of different surface area, Figure 2d. The pressure on the smaller piston will be equal to the pressure on the bigger piston multiplied by the surface area ratio of the pistons. When the needed enhancement factor is large, the bigger piston has to be very big, or the smaller piston has to be very small. A very small piston surface area can be created by using a differential cylinder in which the area of a big cylinder minus the area of a slightly smaller cylinder creates the small piston area.

To turn a pressure intensifier into a two-phase mechanism a switching mechanism is needed, which switches the system between the first (bypass) phase and the second (pressure enhancing) phase. A pressure switch, incorporating a spring loaded piston, has been designed. In its resting position, the fluid can flow freely from the inlet to the lower outlet, Figure 2c. At the moment the pressure increases, a piston is forced to move, closing the lower outlet and opening the upper outlet. To make the switching pressure variable, the pretension in the spring is adjustable.

Curved actuators

Hydraulic actuators are available in several different implementations. Most commonly known are the linear actuators consisting of a cylinder and a piston. Their advantages are a high efficiency and a high maximum working pressure. The disadvantage is that these actuators execute their force in a straight line, which is not always desirable. To enforce a rotary motion one can use a vane motor. Vane motors need more sealing, due to their principle and construction. This will lead to a lower efficiency and higher friction and leakage, which make them unsuitable for use in a prosthesis. In the new VC prosthesis both the master actuator and the slave actuator are connected to a rotating movement. In case of the master actuator this is the wrist joint, and in the case of the slave actuator this is the finger joint. When a linear actuator would be used, an extra link has to be added to convert the linear motion into a rotary motion, as is shown in Figure 3a.

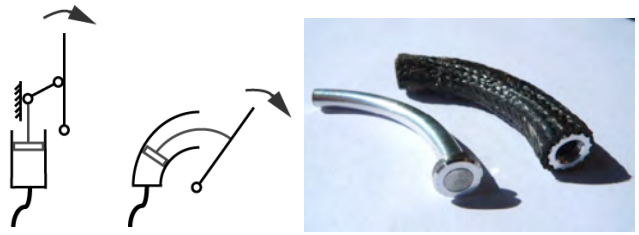

Figure 3: a) Comparison of a straight (left) and curved (right) hydraulic actuator. b) Prototype of the curved actuator.

This extra link will take extra space, increase friction and radial forces on the actuator. With a curved cylinder and piston the extra link could be discarded. Both the piston and cylinder would be a revolved shape around the joint of the connected parts. This would mean a smaller overall size, less friction and no radial forces on the actuator. An added benefit is a linear relation between movement of the actuator and the rotation of the connected joint. Unfortunately, curved actuators are not commercially available. Therefore, these actuators were custom made, Figure 3b. Production of the curved piston is done by producing a standard piston head, which is connected to a bent rod. Production of a curved cylinder however is not so trivial. Machining a curved hole with these dimensions is not possible. Other production

methods, like spark-erosion cavity sinking, would lead to a surface roughness unsuitable for hydraulic use. To overcome these problems, the cylinder is formed from carbon fiber around a male mold. The mold itself can be bent to the right curvature. This will produce a curved cylinder with a surface roughness equal to the male mold used.

EVALUATION

All components are build and tested separately. The pressure intensifier and pressure switch function as expected. The switching pressure can be adjusted in a usable range by adjusting the pretension on the return spring. The production of the curved actuator turned out to be feasible but critical with respect to the tolerances of the cylinder mould. Initially this has led to unacceptable leakage. Several parts of the hydraulic lock are produces by a rapid manufacturing technique. As a result, the tolerances of these parts are not good enough to ensure proper O-ring seating and sealing, and the surface roughness of these parts cause excessive friction, which causes the lock to be not fully functional. Both problems can be solved with a higher print resolution, and/or post printing machining.

CONCLUDING REMARKS

The overall volume of all the components together is small enough to fit in a prosthesis for a four year old child. This is greatly helped by the fact that the relative orientation of the locking mechanism, the pressure switch and the pressure intensifier is not relevant and can be changed to make maximum use of the space available, Figure 4. The mass of all hydraulic components is 25 grams. The amount of hydraulic oil needed is approximately 2 ml, or less than 2 grams. The test results show that it is possible to make a pressure switch and a pressure intensifier, small enough for prosthetic use. The curved actuators show to be functional and capable of reducing building volume in a prosthesis. The automatic hydraulic lock is considered feasible, assuming it is possible to overcome the tolerance and surface roughness problems.

Figure 4: Mock up of a child sized prosthesis (approx. 100mm total length), including all hydraulic components.

Overall, the conclusion is that a hydraulic prosthesis is very well feasible. The mass and size are suitable for (children-) prostheses (see Fig. 21), and the power transfer and efficiency are competitive with traditional mechanical systems. The inherent reliability of a closed hydraulic system is a very big advantage over conventional systems.

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VOLUNTARY CLOSING PROSTHESES

An answer to user needs!

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INTRODUCTION

In order to meet the needs of a person with an arm defect, many, sometimes conflicting, requirements have to be fulfilled. These requirements can be summarized into three basic demands: cosmetics, comfort, and control [1, 2]. In the control domain natural, intuitive, subconscious control is strived for. To achieve this type of control proper feedback needs to be present [3]. This implies control according to the voluntary closing principle. In body-powered, body-controlled prostheses voluntary closing control enables proprioceptive feedback of position, velocity, and force to the user through the bodies own proprioceptive system, comparable to the way we use a tool, e.g. a tennis racket [4, 5].

The Delft Institute of Prosthetics and Orthotics (DIPO) has started several projects to develop body-powered, body-controlled voluntary closing hand prostheses. Current projects include a voluntary closing hand for toddlers, where the reduction of glove forces acting is the main challenge; a voluntary closing prehensor similar in looks to the already existing and successful appealing prehensor [6], where the main challenge is a variable advantage mechanism to promote fast sizing of powerful grasps; a wrist-operated voluntary closing prehensor, where the challenge is to incorporate novel hydraulics; and a study into the psychophysical properties of shoulder harnesses, where the main challenge is to identify the optimal force and excursion windows. Ultimately, we aim at voluntary closing hands that provide adaptive grasp patterns. Adaptivity enhances the natural appearance [cosmetics], and at the same time reduces the need for high pinch forces [comfort, control].

Each of the above mentioned projects is briefly described below.

VC HAND FOR TODDLERS

In active, toddler sized hand prostheses the cosmetic glove introduces a stiffness which causes the required operation forces to be too high to be generated by toddlers. Measurements on several cosmetic gloves of identical size and brand, showed different glove stiffness characteristics. The goal of this project is to design a voluntary closing toddler sized hand prosthesis using an adjustable glove compensation mechanism. A prototype of the glove compensated hand prosthesis was designed. The design is based upon the WILMER passive hand prosthesis for toddlers [7] and utilizes a spring mechanism with negative stiffness to compensate for the glove forces [8]. Future work should determine the feasibility of the design.

VC PREHENSOR

A relatively new project aims to convert the already existing and successful appealing prehensor [6], which operates in a voluntary opening fashion, into a voluntary closing device. Most likely, the design will incorporate a variable advantage mechanism to promote fast sizing of powerful grasps, similar to previous designs made within DIPO [9].

WRIST-OPERATED VC PREHENSOR

Currently, a person with an arm defect can choose between an electrically powered or a body-powered prosthesis. Both options have their own inherent advantages and disadvantages. The electronic prosthesis can be esthetical and easy to use, but is heavy, expensive and vulnerable. The body-powered prosthesis is cheap and reliable, but requires an uncomfortable shoulder harness to be operated.

Over a decade ago, a prototype of a new voluntary closing prosthesis was conceived [10]. It does not need external power or a shoulder harness to be operated, combining advantages of both types of prostheses. This is realized by using passive flexion of the prosthetic wrist to power the prosthesis. An integrated locking mechanism allows the user to hold an object without exerting any operating force.

A recent study [11] showed the feasibility of wrist flexion operation. Subsequently, a prototype of the new voluntary closing prosthesis was designed. It comprises a hydraulic system containing a pressure controlled pressure intensifier, an automatic locking system, and some novel

hydraulic actuators. All components are built and tested. The results show the viability of the new concept [12].

PSYCHOPHYSICAL PROPERTIES OF SHOULDER HARNESSES

High rejection rates indicate users are not satisfied with the performance of their arm prostheses. In theory, the advantage of shoulder controlled prostheses is that the user receives direct proprioceptive feedback about the opening width and pinch force of the terminal device. However, the operating forces of commercially available voluntary closing prostheses are high, leading to discomfort and disturbing the direct proprioceptive feedback.

As a start, a pilot study was performed to find the optimal operation force, at which the user receives optimum force feedback during comfortable prosthesis operation [13, 14]. During experimental research, subjects were asked to reproduce a reference force, with and without visual representation of the forces produced. The subject's performances of blind generated forces regarding the reproducibility, stability and repeatability were evaluated to find an optimal cable force. The performances of male and female subjects, with and without arm defects were compared.

The optimal operation force level is between 20 and 30N for male and female subjects without arm defects. No differences in stability and repeatability performance are found between subjects with and without an arm defect. Subjects with arm defects the reproducibility optimum is found between 10 and 20N as they have difficulties reproducing high force levels (> 30N).

Future work will extend the force measurements and combine them with measurements for the optimal cable excursion feedback. Hence, a proper understanding of the proprioceptive feedback capabilities of a prosthetic user is gained. From here, the optimal control forces and displacements can be determined. These will serve as the basis for the design of a new physiological control system.

VC HAND WITH ADAPTIVE FINGERS

When it comes to body powered prostheses, most users prefer a hook over a hand. Body powered hands require an uncomfortable high activation force [15, 16], produce a relatively low pinch force (<15 N) and have stiff fingers. Despite all its drawbacks, the design of body powered hand prostheses almost has not changed since the 1950's.

The goal of this study was to design a new body-powered, voluntary closing hand prosthesis, which has articulating fingers. This hand should require an operation force within a comfortable level.

A new hand was designed and prototyped [17]. The hand uses hydraulics, to enable an efficient transmission and to avoid the use of an inefficient Bowden-cable. The fingers have articulating MCP- and PIP-joints, enabling both precision and cylinder grasp. The fingers are actuated by individual hydraulic cylinders, which fit inside the fingers. The cylinders can be operated at a high pressure (>50 bar), which enables a high pinch load (>30 N). The user can operate the hand by activating a hydraulic master cylinder, attached to a shoulder strap. The hydraulic hand is fast and reliable, due to the use of body control. The hand provides feedback to the user, which enables accurate force and position control.

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HANDY HOOK REVISITED

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INTRODUCTION

The partial hand device called the Handy Hook which was made by the Robin-Aids Company [1] from the early 1950s to the 1980s and has been unavailable for many years is again available. The Hosmer-Dorrance Corporation [2] is making the attachment kit for this device and it is part #62594. This hook adaptor system is useable for a partial hand to hook attachment as well as a functional split hook adaptor for quadriplegics and other functional loss situations. The adaptor creates a very low profile hook attachment to a hand splint or partial hand socket to provide functional grip.

HISTORY OF THE HANDY HOOK

The Handy Hook was produced by Robin-Aids Company and was part of a line of partial hand prosthetic options they produced and have since discontinued. (Figure 1)

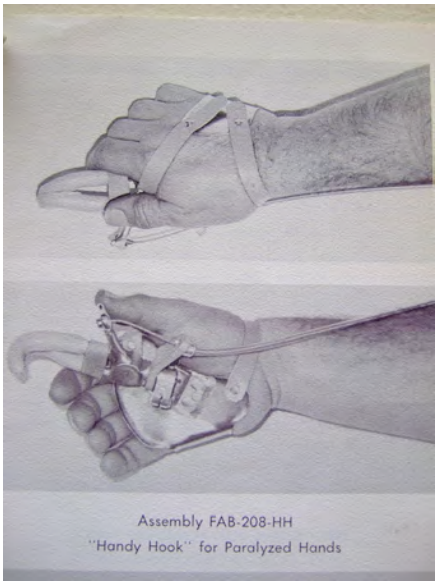

Fig 1 The design of the original handy Hook system discontinued many years ago.

The current design of the hook adaptor (Figure 2) is quite similar to the old design and allows for mounting a split hook on a variety of prosthetic and orthotic devices. The kit consists of parts including a mounting plate with a post that attaches the hook to a threaded friction block. A reaction bar is provided to hold the cable housing to provide opening force to the hook from the split figure-of-eight harness system.

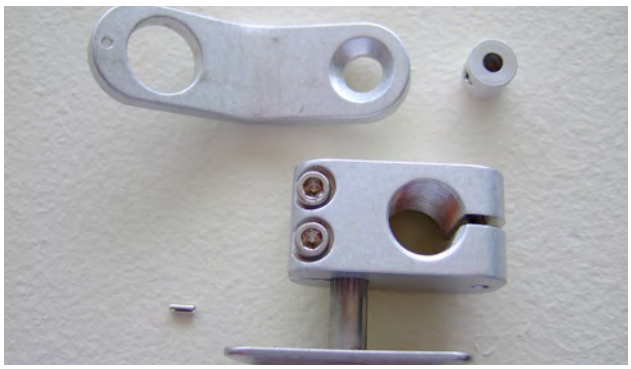

Fig 2 The current parts kit available from Hosmer Dorrance part #62594

This system will provide prehension ability for not only partial hand amputees but is also adaptable to hand splints to provide function in the case of limited or absent hand function. (Figure 3)

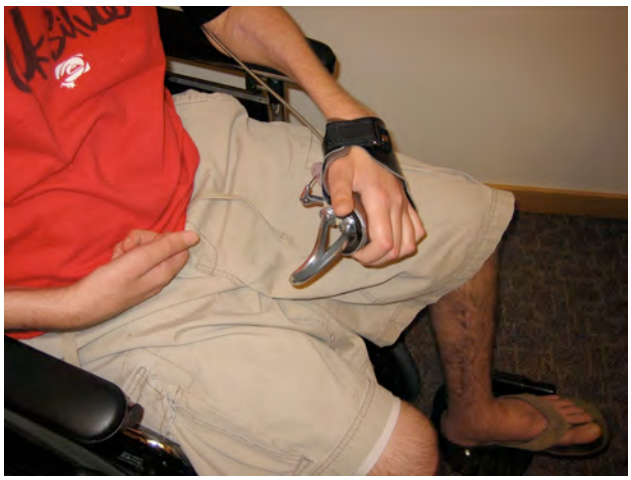

Figure3: Quadriplegic hand splint version of the device

The mounting system can be placed in a variety of locations depending on the needs of the individual. (Figure 4-5) When some hand function is present the hook can be mounted on the back of the hand or close to the wrist joint to improve the length and functional position of the hook.

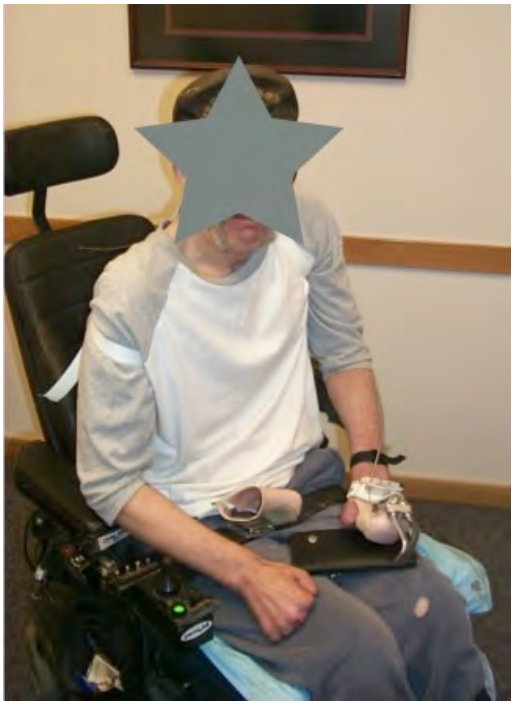

Fig 4 Hook mounted on the back of the hand to allow for partial thumb function without interference

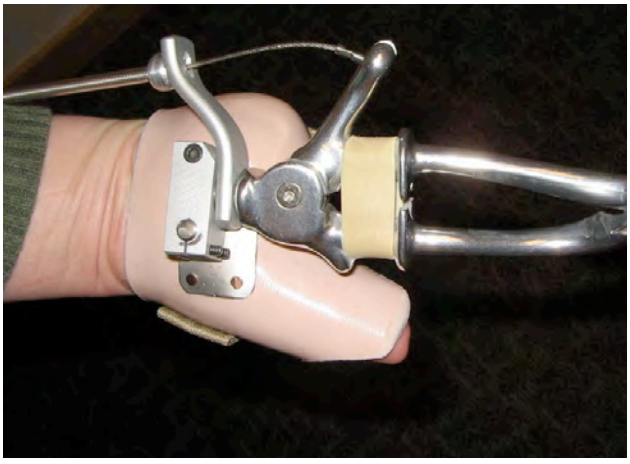

Fig 5 Proximal mounting location to reduce the overall length of the device

The harness system is a traditional split Figure-of –eight design (Figure 6) using scapular protraction to open the terminal device. The cable is easily removable to allow the hook to be used for passive function and positioning when active prehension is not needed.

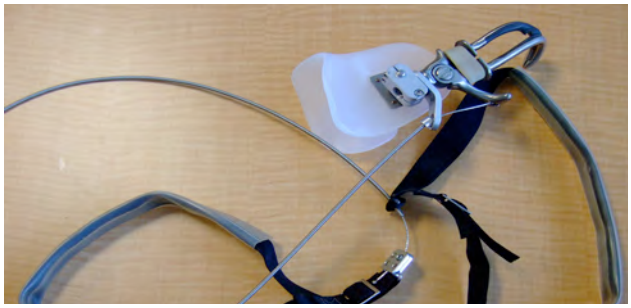

Fig 6 Split harness design

ACKNOWLEDGEMENTS

The author has no financial interest in the Hosmer Dorrance Company and is grateful for their assistance in recreating the Handy Hook kit.

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TOWARD OPTIMIZING ELECTRODE CONFIGURATIONS TO IMPROVE MYOELECTRIC PATTERN RECOGNITION

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ABSTRACT

Standard myoelectric control systems use carefully placed bipolar electrode pairs to provide independent myoelectric signals (MESs) for prosthesis control. Because myoelectric pattern recognition systems do not require isolated MESs, the two electrode poles used for each MES channel may be placed longitudinally along individual muscles or transversely across multiple muscles. In addition, each electrode pole can be combined with a number of additional poles to form multiple channels. However, practical issues limit the number of poles that can be used in clinical settings. In this study, we investigated classification error reduction and controllability improvements provided by a combination of transverse and longitudinal MES channels in two conditions: (1) a constant number of electrode poles, and (2) a constant number of MES channels. In both cases, we also investigated performance when the electrodes were slightly shifted from their original positions to evaluate sensitivity to electrode shift. We found that a combination of two transverse and two longitudinal electrode channels constructed from four poles significantly outperformed the individual performances of either two transverse or two longitudinal channels each constructed from four poles ($p<0.01$). Using eight poles, we found that the best channel subset was always comprised of a combination of transverse and longitudinal channels. These results are important because the number and arrangement of poles and channels is a practical consideration for successful clinical implementation of myoelectric pattern recognition control.

INTRODUCTION

Myoelectric pattern recognition systems show promise for intuitive control of prostheses with multiple degrees of freedom [1]. Despite two decades of extensive research, these systems have yet to be clinically implemented. Typically, studies of pattern recognition systems have focused on signal processing aspects including data windowing [2], feature extraction [3], classification [4], and post-processing [5]. However, a key component of the system is the placement and configuration of the electrode poles. This is true in two contexts: first, the information content of each MES channel is affected by the configuration of the two electrode poles, and second, use of more electrode channels increases computational and financial costs and use of more electrode poles poses technical difficulties in embedding electrodes into prosthetic sockets.

The information content of an MES is determined by the electrode detection volume, which defines the *selectivity* of the electrode. The primary factor affecting selectivity is the interelectrode distance: a rough estimate of detection volume is given by a sphere with radius equal to the interelectrode distance [6]. Most experiments have been conducted with interelectrode distances of approximately 2 cm, resulting in selective recordings. However, in pattern recognition systems, nonselective MES recordings may provide a different set of information that is complementary to the selective information. In addition, nonselective MES recordings may be less sensitive to changes in the location of the recording electrodes [7], such as those that result from donning and doffing the prosthesis or socket shift during use. Electrode shift is a potential problem in clinical applications because data presented to the classifier when electrodes are shifted are different from training data [8].

The number of channels used for classification of MESs varies between studies based on the classification problem and available recording equipment. Four channels are often used in studies of myoelectric pattern recognition control by transradial amputees [9]. One study [4] showed that for a myoelectric control of 10 motion classes using forearm muscles, four channels provided a sufficient amount of information for classification, and additional channels did not increase classification accuracy. In fact, the study showed a small decrease in accuracy as the number of channels was increased to 16. Other studies have shown similar findings with a plateau effect in classification accuracy with increasing numbers of channels [10, 11]. The number of channels is an important property of the MES detection system, and this study examines this property in terms of robustness to electrode shift.

We compared pattern recognition system performance when using combinations of selective and nonselective recordings by measuring classification error and controllability testing. Selective recordings were obtained from bipolar electrode pairs with small interelectrode distances aligned

with the underlying muscle fibers. Nonselective recordings were obtained from bipolar electrode pairs aligned in a transverse orientation to underlying muscle fibers and spanning muscle groups: one electrode pole was placed on the wrist flexor muscle group and one electrode pole is placed on the wrist extensor muscle group. This configuration has a large interelectrode distance and records a global signal from multiple muscles [3].

In this paper, we seek to provide clinical recommendations for electrode placement and number of electrodes and recording channels based on offline classification error, real-time controllability scores, and robustness to electrode shift.

METHODS

Experiment 1: Effect of Pole Number and Placement

Seven able-bodied subjects participated in the study, which was approved by the Northwestern University Institutional Review Board. Two control sites on the forearm were used: one on the flexor muscle group and one on the extensor muscle group. At each control site, two surface electrodes were placed longitudinal to the direction of the underlying muscle fibers. A ground electrode was placed on a bony region near the elbow away from the muscles of interest. Four bipolar MES channels were formed from these four electrode pole locations (Figure 1). Two channels were longitudinal (spanning each individual control site), and two were transverse (one pole on flexors and one on extensors).

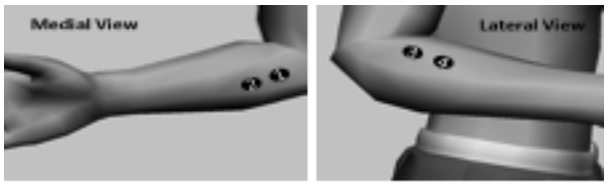

Figure 1: Electrode placement for experiment 1. Electrode poles 1 and 2 are located on wrist flexors and 3 and 4 on wrist extensor muscles. The two longitudinal channels were the bipolar pairs of 1 & 2 and 3 & 4. The transverse channels were 1 & 3 and 2 & 4.

For each subject, the classifier was trained and tested offline with electrodes located at the nominal (or *no-shift*) location. The electrodes were manually shifted 1 and 2 cm from the nominal position in the direction parallel to the underlying muscle fibers (distal to the subject) and 1 and 2 cm from the nominal position perpendicular to the underlying muscles (clockwise from the subject's perspective). Testing data were recorded at each of these four shift locations. Seven motion classes were recorded: wrist flexion, wrist extension, forearm pronation, forearm supination, hand close, hand open, and no movement.

A pattern recognition system similar to that used previously [1] was used to discriminate motion classes. MESs were sampled at 1 kHz and high-pass filtered at 20 Hz. Data were windowed in 250 ms intervals with 50 ms overlap [2]. Time domain features [3] were classified using linear discriminant analysis [1].

The performance of the classifier with three different channel combinations using the same four electrode pole locations were tested: (1) using two longitudinal channels, (2) using two transverse channels, and (3) using a combination of two longitudinal and two transverse channels. We also compared our results to those achieved when data collected from the shifted locations were incorporated into the training data (referred to as *displacement training*) in order to reduce classifier sensitivity to electrode shift [12].

A Target Achievement Control (TAC) test [5] was used to evaluate controllability. Subjects controlled a virtual prosthesis to achieve target postures shown on a screen (see [2, 5] for more details on TAC testing). Only one motion class (e.g. wrist flexion and extension) was required per trial. If mistakes were made, for example, activation of a different motion class or overshooting the target, the subject had to make a corrective activation. The subject had 17 s to complete each trial. A test consisted of two trials of each motion class. Performance was assessed by failure rate: the percentage of trials that the subject did not complete during a test. TAC tests were completed at the no-shift and 2 cm shift (both parallel and perpendicular) locations. At each location, one test was completed for two longitudinal channels, and one test was completed using two longitudinal and two transverse channels.

Experiment 2: Effect of Number of Channels

This experiment was similar to the first except four control sites spaced equally around the circumference of the forearm were used. Eight bipolar MES channels were formed from eight electrode pole locations (Figure 2). Four channels were longitudinal and four were transverse in an arrangement similar to experiment 1.

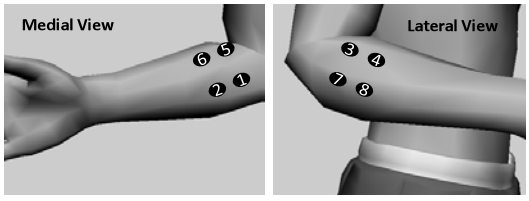

Figure 2: Electrode placement for experiment 2. Longitudinal channels were formed between poles 1 & 2, 3 & 4, 5 & 6, and 7 & 8. Transverse channels were formed between poles 1 & 3, 2 & 4, 5 & 7, and 6 & 8.

The same motions and classification techniques as described for experiment 1 were used. The classifier was trained and tested at the no-shift location. In this experiment, electrodes were only shifted 1 and 2 cm perpendicular to the no-shift location, because results from experiment 1 showed that the classifier was more sensitive to perpendicular shifts than to parallel shifts. Classifier performance resulting from use of from one to eight channels was evaluated and the best channel subset for each number of electrodes was determined based on the error at the no shift, 1 cm shift, and 2 cm shift locations using the following weighted error formula:

*Weighted Error = 2*Error(No Shift) + 1.5*Error(1 cm Shift) +Error(2 cm shift).*

Every combination of channels was tested and a weighted error was assigned to each combination. The optimal combination was that which had the lowest weighted error compared to others with the same number of channels.

RESULTS

Experiment 1: Effect of Pole Number and Placement

In the first experiment, we evaluated the best electrode channel configuration for a constant number of electrode poles. Two longitudinal and two transverse channels were formed from the same four electrode poles in different configurations. The combined selective and nonselective information from both longitudinal and transverse channels performed the best with and without shift (Figure 3) compared to only transverse channels ($p<0.01$) and only longitudinal channels ($p<0.05$).

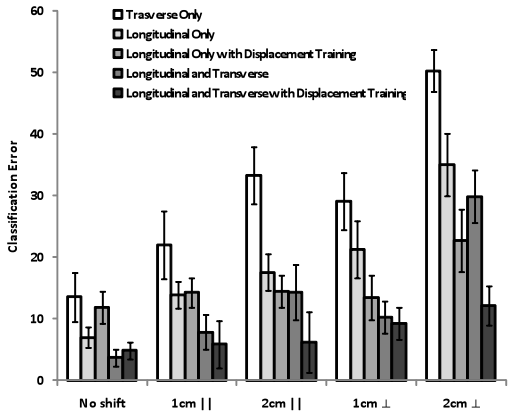

Figure 3: Classification error of electrode configurations using four electrode poles. Displacement training results are also displayed for two of the configurations. Error bars show one standard error of the mean. || refers to parallel shifts and ⊥ refers to perpendicular shifts.

Training with displacement data increased error at the no-shift location in both cases, but also decreased sensitivity to electrode shift (Figure 3). Displacement training did not reduce sensitivity to shift for the longitudinal channels as much as adding the transverse channel did. However, by adding the transverse channels and incorporating displacement training data, sensitivity to shift was greatly reduced at all shift locations ($p<0.05$) compared to a combination of longitudinal and transverse channels without displacement training.

TAC test controllability results demonstrated a trend similar to that of the classification error: a combination of longitudinal and transverse channels outperformed longitudinal channels alone (Figure 4), especially when electrodes were shifted. The controllability test was not hard enough to separate out the performance of the two classifiers at the no shift location, as both had very low failure rates.

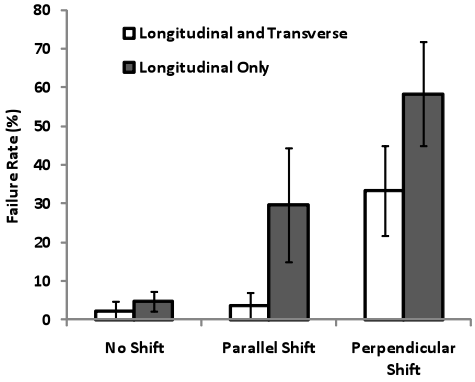

Figure 4: TAC test failure rates for two electrode configurations with and without 2 cm shifts. Error bars show one standard error of the mean.

The longitudinal channels alone performed well without shift, but had high failure rate with shift in either direction. By adding transverse channels, the no-shift failure rate dropped slightly and the failure rate was substantially reduced for both shift directions. In particular, using both longitudinal and transverse channels and a 2 cm parallel shift the failure rate was lower than the longitudinal channels without shift.

Experiment 2: Effect of Number of Channels

First, the best subset of channels for each number of channels (between one and eight) was determined based on their weighted classification errors. Only the unweighted classification errors are displayed here (Figure 5). Interestingly, combinations with at least one longitudinal and one transverse channel were always in the best subset when more than one channel was used.

The use of four to six channels had the lowest classification error across the three shift conditions. An important result was that errors were below 15% at the nonshifted and 1 cm shift locations when using more than two recording channels. The 2 cm shift location had high classification error regardless of the number of recording channels.

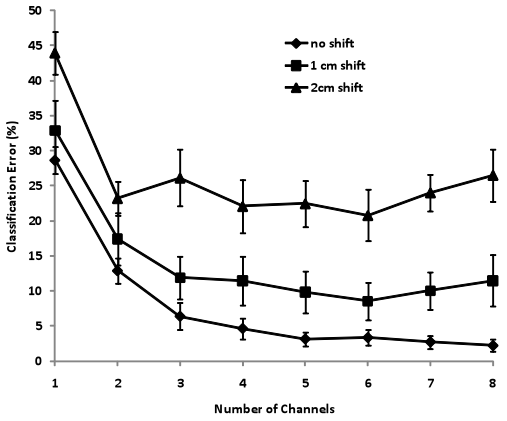

Figure 5: Effect of the number of recording channels on classification error. Error bars show one standard error of the mean.

DISCUSSION

The first goal of this study was to find the electrode configuration that gave the highest performance given a limited number of electrode pole locations. Performance was measured in terms of classification error, controllability, and robustness to electrode shift. We found that a combined configuration that included both longitudinal and transverse channels was the highest performing configuration of those that were tested. This configuration had lower error and better controllability without shift and at every shift location tested compared to using only longitudinal or transverse channels.

Previous investigators have considered including displacement location data in the training data in order to train a more robust pattern recognition system [12]. We repeated this analysis for the configuration with longitudinal channels and the configuration with combined longitudinal and transverse channels. We found that displacement training increased classification error at the no-shift location, but helped to reduce sensitivity at shift locations. The combination of displacement location with longitudinal and transverse channels performed especially well with less than 15% classification error at all tested locations. This is a clinically useful result as with only four channels and four pole locations, low classification errors were achieved with high robustness across shift conditions.

The second goal of this study was to analyze the number of recording channels necessary for sufficiently high classification and to determine the optimal composition of channel subsets. Based on weighted averages, it was found that a combination of longitudinal and transverse channels was always optimal regardless of the number of channels. The combination of selective and nonselective information decreased error with and without shift compared to configurations with only one type of electrode configuration. Experiment 2 showed that there was little or no improvement in terms of classification error when using more than six channels. For the transradial case, this study demonstrates that four to six channels are sufficient to obtain classification errors of less than 15% both without electrode shift and with shifts up to 1 cm.

ACKNOWLEDGEMENTS

The authors would like to acknowledge Ann Barlow and Aimee Schultz for helping to revise and edit the manuscript, Annie Simon for providing valuable contributions, and Todd Kuiken for providing overall project guidance for this study.

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A NOVEL RESEARCH AND CLINICAL APPROACH TO USING GEL LINERS FOR COLLECTION OF SURFACE MYOELECTRIC SIGNALS FOR PROSTHETIC CONTROL

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ABSTRACT

For more than two decades, individuals with lower limb amputations have been successfully fitted with gel liners constructed from a variety of materials. Prosthetists have also reported moderate success with gel liners fit to individuals with upper limb amputations who use externally powered prostheses. At the Center for Bionic Medicine, we have explored a novel approach to collecting myoelectric signals from individuals with lower limb or upper limb amputations—using electrodes embedded in gel liners. Initial designs have proven more comfortable and easier to don than traditional suction sockets and have allowed us to eliminate the need for separate connection of pre-amplifiers. We believe this technology will be of benefit to individuals with upper or lower limb amputations and eliminate some of the clinical challenges and reported drawbacks of current myoelectric fittings. The next step is to combine the new liner technology with advanced electronics to control actuated drive units in both upper limb and lower limb prostheses. In this contribution we describe the evolution of this liner technology from initial experiences through current status to future directions.

INTRODUCTION

Suction Sockets

Many prosthetists have used suction suspension as the primary or sole means of suspending transhumeral or transradial prostheses. Traditional suction sockets, where the limb is in direct contact with an undersized socket, have been widely used, particularly in the transhumeral population. Much of the theory behind the design of these sockets comes from experiences fitting individuals with transfemoral amputations. Traditional suction sockets provide all of the benefits of total contact sockets, including distribution of forces over larger surface areas to decrease concentrated areas of pressure, decreased edema, increased control of the prosthesis, and enhanced proprioception of the terminal end of the prosthesis [1, 2]. Additional benefits to upper limb–prosthesis users may be an increased abduction range of motion (ROM) and better cosmesis, both due to lower lateral trim lines. Elimination of the harness is also possible depending upon which input devices are used and whether or not a hybrid system incorporating body-powered components is used. At a minimum, adding suction to the total contact socket concept has enabled many individuals to tolerate the use of transhumeral prostheses by decreasing the amount of pressure in the contralateral axilla [3], thus preventing neuropathies of the contralateral arm and hand [4]. In a case study, Vacek [5] reported that elimination of tight harnessing prevented tingling or sensation loss, and concluded that prosthesis comfort directly affects an individual's tolerance of, and desire to continue to wear, the device.

Quasi-Hydrostatic Fittings

If the socket is appropriately undersized, it offers the ability to achieve a semi- or quasi-hydrostatic socket environment. Stokosa [6] coined the phrase *Total Surface Bearing (TSB)* for transtibial socket design in which "*the entire surface of the residual limb is in total contact with (the) socket while every unit area is under compression to its proportionate tolerable level,*" and points out that there is a difference between TSB and hydrostatic concepts. Kahle [7], comparing transtibial designs, defines hydrostatic design based on the mechanical properties of fluids and Long [8] states that utilizing the hydrostatic interface design promotes tissue elongation, increasing distal padding and producing a residual limb with a firmer tissue consistency. This is especially important for control of the prosthetic socket in levels of amputation where the soft tissue is less more mobile, which occurs when only one bone is present in the soft tissue—for example in transhumeral or transfemoral amputations. Kahle [7] contends that a proximal seal with the humeral epicondyles is essential for achieving a hydrostatic fit when utilizing roll-on gel liner technology in transtibial sockets. Miguelez [9] asserts that a secondary benefit of hydrostatic fit— the lack of movement of transradial sockets during loading—can be attributed to muscle contouring as opposed to soft tissue compression alone. If true, these latter two statements suggest that liners must be designed so as to (i) create a seal at the proximal socket (gleno-humeral joint or humeral epicondyles for transhumeral or transradial limbs, respectively) and (ii) capture more muscle contours.

The method used to don the socket will affect the ability to achieve a hydrostatic fit. In traditional suction sockets, hydrostatic fit has been achieved with the use of a donning aid. This donning method is challenging for individuals with unilateral and bilateral amputations as they need to balance their prosthesis while simultaneously donning the device. Additionally, the inner socket is customarily made of a semi-rigid or rigid thermoplastic, which may become uncomfortable as the user attempts to attain the end ranges of shoulder motion (i.e. glenohumeral flexion and abduction). This is due to the weight of the device creating a force couple that places an intolerable pressure on the distal aspect of the humerus.

The first four individuals with transhumeral amputations who underwent targeted muscle reinnervation (TMR) were fit with traditional suction sockets and cited the donning method as one of the major deterrents to wearing the device: an additional difficulty for TMR subjects is the necessity of precisely orienting their limb with respect to the electrode contacts within the prosthesis in order to achieve optimal alignment for myoelectric control.

Gel Liners

Early designs of gel liners were custom-fabricated over a modified positive model. Ossur Kristinsson first developed this technology, which evolved into viable off-the-shelf liners. A majority of residual limbs can be fit well with appropriately sized off-the-shelf liners, although custom-fabricated liners are still utilized for limbs requiring special attention. Most gel liners are fit to individuals with lower limb amputations for reasons of comfort, suspension, and because the increased shear force between the limb and liner (and decreased shear between the liner and socket) protects skin on the residual limb from friction caused by relative movement of limb and socket interface. An additional benefit is the option of applying sub-atmospheric pressure to the limb-socket interface. Some gel-liner manufacturers use terms such as *TSB* and *hydrostatic* in their product information, however, these varied fitting goals are achieved in many different ways and, although some are based on published specifications, are quite generic in their product applications. Attention should be paid to using these liners as hydrostatic fittings as this requires a distal distraction of the residual limb, which creates elongation and a reduction in cross-sectional area. One method of employing this change in soft tissue geometry is with the use of a lanyard added to the end of the liner to pull the limb into the socket. Another technique, more easily implemented in lower limb prostheses, is using a liner with an added distal pin: while the pin is engaged in the locking mechanism, repetitive loading (weight bearing) and unloading will elongate and circumferentially reduce the limb in size and the pin will further engage into the locking mechanism.

Although roll-on gel liners have been historically used in lower limb fittings, there has been some previous use of this technology with upper limb prostheses. Radocy [10], who has a transradial amputation, presented some of the earliest information on, and evaluation of efficacy of, roll-on gel liners with upper limb prostheses. Early in the development of *silicone suction socket (3S)* technology, it was reported that this fit prevents pistoning of the prosthesis and reduces or eliminates perspiration because there is no air layer between the skin and the socket wall [10]. Radocy [10] reported that the combination of a supracondylar socket and silicone liner provided superior suspension, improved performance during rigorous activities, and reduced or eliminated residual limb–to–socket rotation. The reduction in pistoning and rotation is beneficial for both suspension and maintenance of skin integrity; however, lack of perspiration may decrease surface myoelectric signals.

Daly [11] and Salam [12] utilized roll-on gel liners for both transhumeral and transradial fittings using different techniques for myoelectric signal detection.

In each study, when using roll-on gel liners in transhumeral or transradial sockets, individuals were able to achieve increased ROM. In addition, lower trim lines were possible. For transradial subjects, Daly [11] reported an average increase in ROM of 22.33º and an increase of pull force (before losing suspension) of 30 lbf. Daly [11] reported an average ROM from 8.57º to 120º and a pull force of 37 lbf (with two of the trials exceeding 50 lbf) for his transhumeral subjects. Since no comparisons were made between higher trim lines and roll-on gel liners, it is difficult to determine whether the lower trim lines or the gel liners caused the increase in ROM. Miguelez [9] might argue that neither is the determining factor, due to the fact that his transradial socket design has neither lower trim lines than conventional fittings nor does it necessarily utilize gel liners, yet he reports greater range ROM.

Salam [12] cut holes in gel liners to allow skin to protrude and make contact with the electrodes, and Bill Hansen (Liberating Technologies, Inc.) has proposed using gel liners with conductive patches. Both approaches have similar drawbacks: the hole/patch location must be exactly placed and cannot be moved. However, both have the potential benefit that all wires are self-contained. Daly provided for the transmission of myoelectric signals through the gel liners via snap electrodes—a method we commonly employ in a research setting. This technique permits the individual to don the liner (with contacts incorporated) and then snap a wire harness to the electrodes before inserting the liner into the socket. One major drawback with this approach is the need to protect the wire harness. Both Daly and Salam claim that users are able to don the liner so that the electrodes consistently

end up in their correct location. This can be accomplished by practice, by referencing anatomical landmarks, or by referencing marks tattooed on the skin [12]. Salam described the ability to have "more proximal placement of electrodes, if needed, without fear of breaking suction" [12]; his conclusion is based on how the electrodes contact the skin and the size of the electrodes and pre-amplifiers. The underlying question to answer is, is one method of socket interface more reliable or repeatable for electrode placement and congruity than another?

DISCUSSION

The evolution of our gel liner design has involved many changes. Much of the earlier work investigated using a stainless steel contact dome and custom-fabricated stainless steel discs (buttons) to create electrode contacts. These configurations would transmit myoelectric signals through the liner and form a junction with a disc magnet that was embedded in the inner wall of the socket and attached to an external wire leading to the pre-amplifier (Figure 1). This method works well in the laboratory setting on able-bodied subjects and for some individuals with transradial and transhumeral amputation, but requires further refinement and investigation. Challenges arise when the prosthesis user experiences significant movement of residual limb soft tissue so that the button disengages from the magnet. This was problematic for subjects with transhumeral amputations who had undergone TMR and had substantial movement of their soft tissue during muscle contractions.

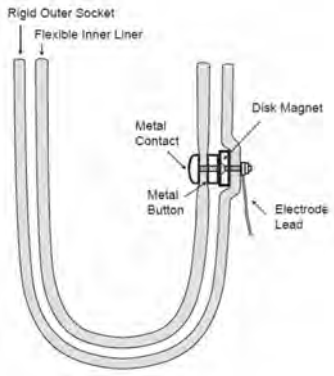

Figure 1: An example of an early iteration of electrode/signal contact interface between gel liner and socket

While investigating new socket interface designs for myoelectric fittings, we became involved in the DARPA Revolutionizing Prosthetics 2009 project. Specifications outlined there steered us toward redesign so that the myoelectric signals were fed into the electronics at the distal aspect of the limb-socket interface. It was necessary to transmit the myoelectric signals along the liner to the distal end through wires or conductive leads. Since the liner was to be inverted and rolled 180° with respect to itself, the signal transmitting material used had to be flexible enough to withstand severe and repetitive flexing. We have investigated using a conductive textile fabric in an attempt to create a liner with signal transmission leads that can undergo the donning and doffing process without serious fatiguing or failure, and would ideally maintain myoelectric signal quality and continuity throughout the useful lifetime of the liner. In order to use this material, custom distal connectors had to be fabricated to receive the leads as they exited the liner. The fabric leads used with these liners are currently being investigated. Various sizes, shapes, and durometers of contacts are being optimized through an experimental process to determine the optimal design thus far; this concept has evolved through numerous designs and has been used in trial fittings. These experimental liner systems have been tested in conjunction with the latest electronic hardware and software developments at the Center for Bionic Medicine (CBM) at the Rehabilitation Institute of Chicago.

FUTURE WORK

Much of the preliminary fitting of these liner systems has been within the research setting at the CBM at the Rehabilitation Institute of Chicago. Refinement of the liner design and interface continue as these systems need to be robust enough for field testing. It is our hope that such testing will provide valuable feedback regarding durability and effectiveness of these liners in a real-world setting.

ACKLOWEDGEMENTS

This work was supported by the Defense Advanced Research Project Agency (DARPA) project BAA05-19, US Army Medical Research and Materiel Command project W81XWH-10-2-0033 and US Army Telemedicine and Advanced Technology Research Center (TATRC) project W81XWH-09-2-0020.

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VIRTUAL REALITY SIMULATOR FOR TRAINING AND EVALUATING MYOELECTRIC USERS

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ABSTRACT

Controlling multiple degrees of freedom intuitively and efficiently is a major goal in the field of upper limb prosthetics. Many novel algorithms have been conceived to meet this goal, but few have been tested in a functionally relevant manner. We have developed a virtual reality myoelectric prosthesis simulator for testing novel control algorithms and devices. The system acquires EMG commands and residual limb kinematics, simulates the prosthesis dynamics, and displays the combined residual limb and virtual prosthesis movements in a virtual reality environment that includes force-based interactions with virtual objects. Both a transhumeral and transradial simulator have been developed. The transradial simulator includes a dexterous hand and haptic feedback to the residual limb. The virtual reality prosthesis simulator is a promising tool for evaluating control methods, prototyping novel prostheses, and training amputees. Due to its relatively inexpensive and portable components (excluding the haptic device), the simulator can be used in the lab, clinic, or at home.

INTRODUCTION

Researchers have sought to improve control of myoelectric prostheses for several decades. A major goal is to achieve simultaneous, or at least seamless, control of several degrees of freedom. Many promising control algorithms have been developed using pattern recognition techniques. However, few of these algorithms have been tested in a closed-loop manner, and even fewer have been tested in a functionally relevant way. Virtual reality has been suggested as a method to quickly develop and evaluate control strategies, prototype devices, and train subjects.[1-3]

Previous myoelectric prosthesis simulators have included costly or complicated components making them impractical for widespread use in clinical settings [1,2]. Also, only recently have advances in computer hardware and development of real-time physics simulation software—driven by widespread use in commercial video games—made real-time simulation of many physical interactions a possibility.

We recently developed a transhumeral simulator [3] and demonstrated a standard clinical assessment within the virtual reality environment, using a force-based physics engine. We have now developed a dextrous hand for transradial simulations and added the capability for haptic (i.e. force/touch) feedback to simulate collisions and inertial affects of the prosthesis.

METHODS

The simulator user, with or without upper limb loss, views an animation of his or her residual limb movement and the simulated prosthesis movement in a virtual environment. The system includes several components: kinematic tracking, EMG or command acquisition, data analysis and control, physics simulation, visualization, and haptic feedback. These components are illustrated in Figure 1 and described in detail below.

1.Kinematic Tracking

Kinematic tracking of the residual limb can be achieved in multiple ways. For a portable system, one or more orientation sensors (e.g., 3DM-GX1, MicroStrain Inc., Williston, VT) are used to accurately measure the orientation of limb segments by fusing signals from triaxial accelerometers, gyroscopes, and magnetometers. Since the sensor always outputs heading relative to magnetic North, the subject should keep his or her body oriented in the same direction during trials. Since the subject should always face the computer monitor, his or her direction with respect to North can be calibrated quickly using the same sensor. If motions of two limb segments (e.g. upper and fore arm) need to be recorded, two sensors are required. Otherwise one sensor is sufficient.

In a lab setting, the HapticMaster[4] (Moog FCS, Netherlands) can be used to record the 3D position and the 3D orientation of the residual limb segment via a 3-dof instrumented gimbal. Since the position returned by the HapticMaster is the proximal attachment point of the gimbal, some simple forward kinematic calculations based on the segment lengths of the gimbal are required to get the position of the end of the residual limb segment.

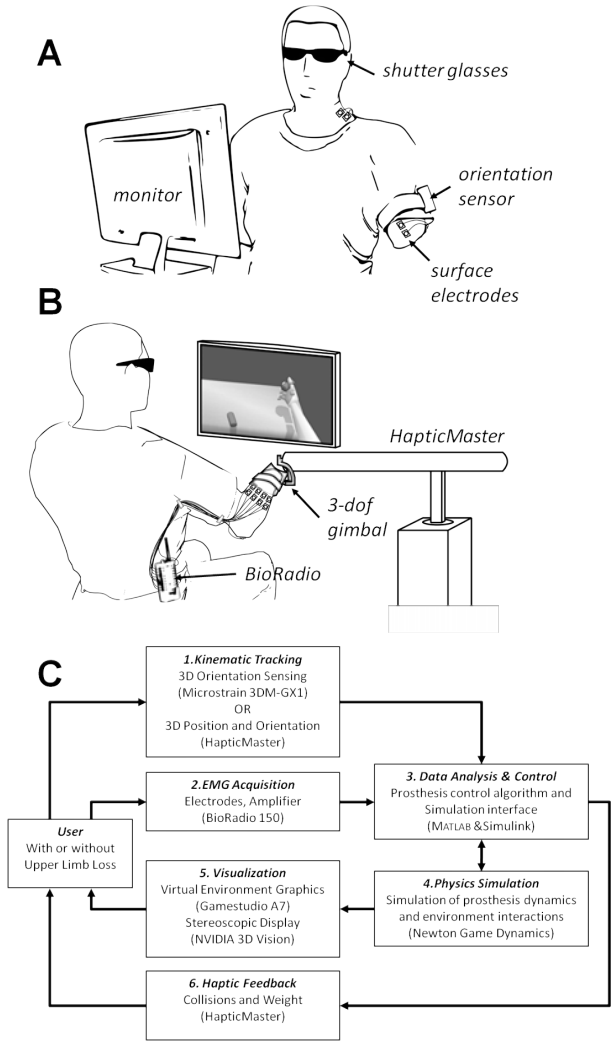

Figure 1: (A) Portable simulator setup for a transhumeral amputee. (B) Lab-based simulator with haptic feedback for a transradial amputee. (C) Flow chart for simulator. Each component is described in detail in the text.

2.EMG Acquisition

Electromyogram (EMG) signals are amplified and collected using an 8-channel wireless data acquisition system, the BioRadio 150 (Cleveland Medical Devices Inc., Cleveland, OH), and disposable snap-type surface electrodes. The BioRadio samples up to 960Hz. EMG filtering and feature extraction is done in software (see next section). The device also has an auxiliary input that can be used to interface with non-EMG based prosthesis inputs (switches, linear potentiometers, etc.).

3.Data Analysis & Control

The HapticMaster, orientation sensor, and the BioRadio as well as the visualization system (described in the next section) are interfaced with MATLAB & Simulink (The MathWorks Inc, Natick, MA). Use of the Simulink block diagram interface results in a modular system that is very easy to customize. For instance, the system allows for almost unlimited customizability in EMG processing and command algorithms. Time-domain feature extraction methods [5] commonly used in pattern-recognition systems have been implemented into the simulator. Also digital high-pass filters for eliminating motion-artifact and band-pass filters for eliminating 60-Hz noise are implemented.

4.Physics Simulation

Physics simulation is implemented using Newton Game Dynamics (NGD, newtondynamics.com), a deterministic force-based solver, used in previous hand simulations [3,6]. Table 1 summarizes the dofs currently modelled in our simulators. The virtual hand prosthesis used in the simulator has individually compliant "motorized" digits that are controlled in concert, similar to currently available dexterous hands (i-LIMB, BeBionic, and Michelangelo).

Table 1: Prosthesis functions in simulators

Transhumeral (based on Utah Arm 3)	**Transradial (based on dextrous hand)**
• elbow flexion/extension • wrist pronation/supination • hand opening/closing	• wrist flexion/extension • wrist ulnar/radial deviation • thumb palmar abduction/adduction • hand opening/closing

Each prosthesis segment is described by a collision hull that accurately matches the shape of the segment (including concavity), an inertial matrix and mass—affects dynamic properties, and friction and elasticity of its surface—affects its interaction on other surfaces. The prosthesis dofs are described by stiffness, joint limits, and maximal torque to match a desired joint angular velocity.

Objects in the environment are treated as rigid bodies and can take any shape. Varying frictional coefficients and masses can make objects easier or more difficult to grasp.

The residual limb segments are kinematically constrained to match the user's actual residual limb segments. Because obstacles exist in the virtual environment but do not exist in reality, this kinematic matching can result in a "paradoxical" situation in which the physics engine cannot solve for all the constraints in the system. The residual limb constraints are less stiff than the virtual prosthesis joint constraints, such that the shoulder will "dislocate" slightly allowing the prosthesis constraints to be maintained in these situations. Still, despite this, and simulated tasks being carefully designed to minimize collisions, some user education is required. Alternatively a haptic feedback device can be used to actually move the user's arm away from the collision (see section below).

5.Visualization

The visualization is implemented using a custom application made with Gamestudio A7 game development system (Conitec Datasystems Inc., La Mesa, CA). This game engine is highly flexible, but easy to use, and supports soft skin deformation, dynamic shadows, high-quality 3D graphics, and custom plug-ins. The physics simulation described above is also incorporated into this application. Kinematic data from the residual limb and prosthesis commands are received from Matlab through a custom plug-in.

6.Haptic Feedback

The HapticMaster can be used to apply forces to move the user's residual limb and simulate the weight and inertial effects of a prosthesis. Force vectors from collision points in the NGD physics simulations can be queried and sent to the HapticMaster, via the Matlab interface, and applied to the user. In this manner, the user actually feels collisions caused by their movements in the virtual environment. The actual force applied by the user is measured by the onboard force sensors. The applied "reaction" force can thus be normalized to the user's force.

RESULTS

Figure 2 shows a screen capture from the transhumeral simulator, demonstrating various objects that can be manipulated in the workspace. Figure 3 shows several screenshots from the transradial simulator, highlighting the dextrous capability of the hand. Note that the fingers are simulated with two segments rather than three. Currently available dextrous hand prostheses also use two-segment fingers (i.e. the distal phalangeal joint is fused)

The combination of 3D graphics, immersive stereoscopic viewing, accurate dynamics and collision simulation make the simulator quite compelling. Preliminary users do not need much time to become accustomed to operating the virtual prosthesis in the virtual environment.

DISCUSSION

Implications

In the simulator, kinematic recording of the residual limb is required because most manual tasks require positioning the whole arm, not just the joints of the prosthesis. Positioning of the residual limb also has other implications as decoding algorithms can be highly dependent on posture [7]. For instance, the biceps and triceps—commonly used as command sources in transhumeral prostheses—are both biarticular muscles that normally cross both the elbow and shoulder joint and therefore may become active when the shoulder is moved, possibly eliciting unintended commands. Therefore, it is necessary to test control algorithms under various postural conditions to insure that they will be robust during regular use. The prosthesis simulator is an ideal tool for evaluating of control algorithms in all postures for both users with and without limb loss.

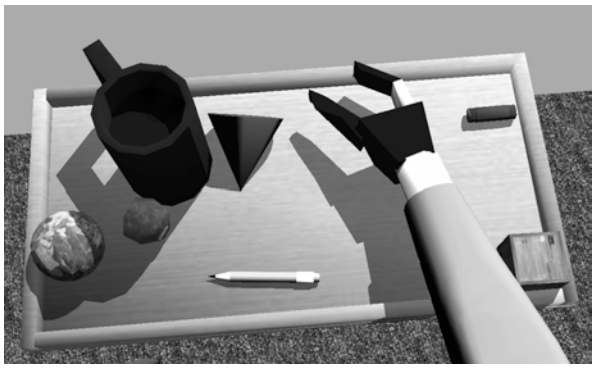

Figure 2: Screen captures from the transhumeral simulator. The terminal device is shown without a glove.

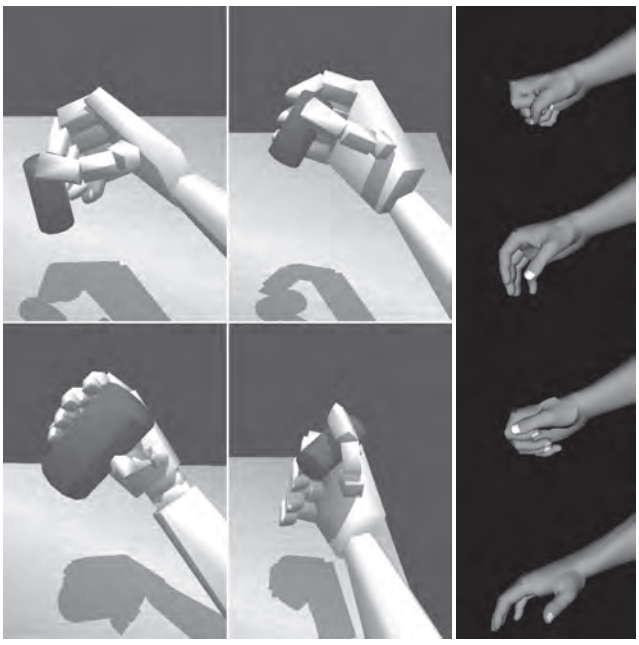

Figure 3: Screen captures from our prototype dextrous hand simulator, showing several objects and grasp types. The final system will hide collision hulls (left) from the user and display only a realistic model of the hand or simulated prosthesis (right)

The simulator can also be used to test novel devices before they are manufactured. For, instance our simulated dextrous hand includes active flexion/extension and ulnar/radial deviation, which is not available in any commercial prosthesis system. Furthermore, maximum speeds, friction, masses can all be adjusted to test their influence on performance. The simulator is a cost effective method for trying new device concepts.

Finally, the simulator can be used for myoelectric training and evaluation. We have previously demonstrated a "Box and Block Test" [8] with the transhumeral simulator. In the study two command methods were evaluated. Using a more traditional command method, normally-limbed subjects moved, on average, the same number of blocks as real amputees performing the same, but non-virtual task[3].

Limitations and Future Work

The HapticMaster can only apply forces from one location and cannot generate torques on the gimbals. Thus the resulting movement of the real and virtual arm may not match upon a collision. However, the haptic feedback was still found useful for avoiding paradoxical situations described above.

Orientation sensors are dependent on a constant "recorded North" direction in the workspace. Large ferrous objects (e.g., filing cabinets, lab equipment) can affect the local magnetic field quite drastically, resulting in inaccurate orientation measurements. Care must be taken when setting up the simulator to insure that these objects are far enough away.

We have demonstrated the ability to grab and manipulate varied rigid bodies in the virtual environment, and developed a virtual Box and Block Test. Future work could include developing additional clinical tests of hand/arm dexterity and function.

CONCLUSION

We have developed virtual reality simulators for training and evaluating myoelectric users in a functionally relevant manner. Virtual prosthesis dynamics and interactions with objects in the environment are simulated using a real-time physics engine. The simulator allows for customization of the prosthesis properties, EMG processing techniques, and the command and control methods. Haptic feedback is useful for more realistically simulating the task. However, without haptic feedback, the simulator is portable, easy to setup, and relatively inexpensive, allowing for widespread clinical use.

ACKNOWLEDGEMENTS

Funding support from TATRC W81XWH-07-2-0044.

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Thursday, 17 August 2011

Wu Conference Centre Auditorium

8:30 AM	**KEYNOTE - Jonathan Wolpaw**

Presentation of Papers

Time	Paper Title	Presenter
9:30 AM	Forequarter Amputee and Mirror Therapy; A Case Report on Adapting the Mirror Box Design	Vera G. van Heijningen
9:45 AM	Long Term Results of Early Myoelectric Fittings	Liselotte Norling Hermansson
10:00 AM	***REFRESHMENT BREAK / EXHIBITOR SHOWCASE***	
10:30 AM	Continuous and Simultaneous EMG-Based Neural Network Control of Transradial Prostheses	Christopher Pulliam
10:45 AM	Sensor Options and Control Strategies for Multi-articulating Hand and Partial-hand-prostheses	Stefan Schulz
11:00 AM	Unique Situations in Prosthetic Design when Applying Targeted Muscle Reinnervation in Transhumeral and Shoulder Disarticulation Levels of Limb Loss	Pat Prigge
11:15 AM	Methods for Collecting EMG from Individuals with Lower Limb Amputations	Robert Lipschutz
11:30 AM	Real-Time Recognition of User Intent for Neural Control of Artificial Legs	Fan Zhang
11:45 AM	Myoelectric Control of a Powered Transfemoral Prosthesis during Non-weight Bearing Activities	Levi Hargrove
12:00 PM	***LUNCH BREAK***	
1:00 PM	Motor Control Processes when Learning to use a Prosthetic Device	Raoul Bongers
1:30 PM	The Effect of Visual Biofeedback Forms of Myoelectric Signal on Syoelectric Signal Separation Training	Kengo Ohnishi

Thursday, 17 August 2011

Presentation of Papers (Continued)

Time	Paper Title	Presenter
1:45 PM	Training Individuals to use Pattern Recognition to Control an Upper Limb Prosthesis	Kathy Stubblefield
2:00 PM	Functioning of Children with Unilateral Congenital Below Elbow Deficiency: An Online Gocus Group Study	Corry K. van der Sluis
2:15 PM	Upper Limb Prosthetics Services Post Haiti Earthquake	Colleen O'Connell
2:30 PM	Designing for Affordability, Application and Performance: The International Trans-radial Adjustable Limb (ITAL) Prosthesis	Alwyn Johnson
2:45 PM	Occupational Therapy for a Multiple Limb Loss Military Patient, a Case Study	Joseph Butkus
3:00 PM	***REFRESHMENT BREAK / POSTER EXHIBITION***	
3:30 PM	Using Multiple Outcome Measures to Determine Skill Level in Myoelectric Prosthesis Use	Hanneke Bouwsema
3:45 PM	Outcome Measure Results of a Pattern Recognition Control of Multi-function Hand Wrist System: A Case Study	Laura Miller
4:00 PM	A Preliminary Study of Learning to Use a Trans-radial Upper Limb Myoelectric Prosthesis	Mohammad Sobuh
4:15 PM	Motion Analysis to Measure Outcomes following Targeted Muscle Reinnervation Surgery	Jacqueline S. Hebert
4:30 PM	Using Motion Analysis to Augment Upper-limb Prosthetics Outcome Measures	Craig Heckathorne
4:45 PM	Use of a Dynamic Load Strap in Adjustable Anatomical Suspension in Transradial Amputee	Sam L. Phillips
5:00 PM	Comprehensive Arm Prosthesis and Rehabilitation Outcomes Questionnaire (CAPROQ)	Shawn Swanson Johnson

KEYNOTE:

THE FUTURE OF BRAIN-COMPUTER INTERFACES: MEETING THE EXPECTATIONS

Jonathan R. Wolpaw

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ABSTRACT

BCIs have a promising future, with researchers in laboratories all over the world using many different brain signals, recording methods, and signal processing approaches to realize increasingly capable systems. These BCI systems can control a variety of external devices, from cursors and avatars on computer screens, to televisions and wheelchairs, to robotic arms and neuroprostheses. People with and without disabilities have tested these systems, and a few are already using them for important purposes in their daily lives. Thus, BCIs are poised to become a major new technology for people with disabilities, and possibly for the general population as well. Nevertheless, the realization of this bright future depends on advances in four critical areas. First, both non-invasive and invasive BCIs need better signal-acquisition hardware. Second, the real-life usefulness of BCI systems for people with disabilities requires convincing clinical validation. Third, effective strategies for BCI dissemination and ongoing support must be developed. Fourth, and perhaps most important, if non-invasive or invasive BCIs are to be widely used for anything more than the most basic communication functions, their reliability must be greatly improved. The difficult problem of reliability may require BCI design strategies based on the principles underlying the excellent reliability of natural neuromuscular actions. These strategies include: effective engagement of brain adaptive capacities; task-appropriate distribution of control between the brain and the BCI; and BCI use of signals from multiple brain areas.

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FOREQUARTER AMPUTEE AND MIRROR THERAPY;

A case report on adapting the mirror box design

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Ruud W. Selles[1] PhD, Wim G.M. Janssen[1]MD PhD

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INTRODUCTION

Mirror therapy has become a valuable treatment for persons with phantom pain. This presentation describes a case study exploring the possibilities of mirror therapy in patients with a forequarter amputation.

AIM

To realize mirror therapy for persons with a forequarter amputee with severe phantom pain.

BACKGROUND

In September 2008, Mrs. M. underwent a forequarter amputation on her right body side, secondary to the recurrence of a mamma carcinoma. In October 2009, she visited our rehabilitation department with complaints of phantom limb pain. Based on this, we decided to start a trial to explore mirror therapy in this patient.

PROCESS

The standard mirror did not successfully create the illusion of the amputated limb being present using the reflection of the unamputated side. Table size mirror was too small to fit the whole arm and shoulder. A large "dressing" mirror was able to solve this problem for the whole arm. However, the reflection of the shoulder and arm was disturbed by mirroring due to the high amputation level in this patient, leading to the visibility of a part of the amputated upper limb during the mirror therapy. For a good illusion of the amputated limb, the unamputated limb needed to be completely blinded for the patient during the mirror therapy. Therefore, a new mirror design had to be realized.

Criteria for function and design:

→ Reflection of the unamputated arm and shoulder

→ Blinding the unamputated arm and shoulder

→ Usable for right and left side amputees

→ Adjustable for tall and small people

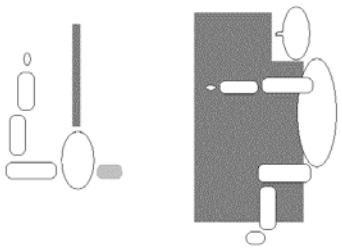
→ Stored easily

→ Movable and transportable easily

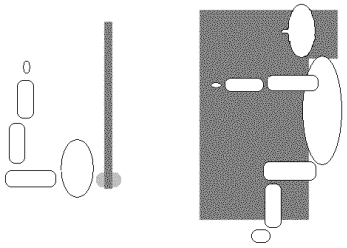
→ To carry along and forward easily by one handed person

We used the following materials: lightweight aluminum for the frame, Perspex mirror sheet and Perspex white sheet for the mirror box. The advantages of Perspex are that it is light weight and easily shaped in the desired design.

We tried two designs:



Design 1



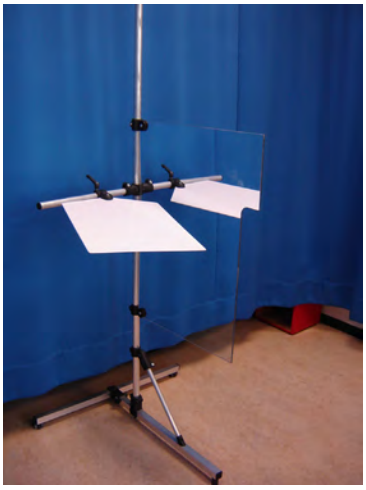
Design 2

Design 1: the amputated shoulder site was still visible, or the patient was unable to resist the urge to move his head to the amputated side.

Design 2: the amputated shoulder site is visual and physical blocked with the design

Two sheets of Mirror Perspex were glued together and assembled with the aluminum frame into the mirror box. A small sheet of white Perspex was used to blind the unamputated arm.

Design 2 fitted the criteria, so this was further developed. The final design is easily adjustable from left to right sided and vice versa, this will takes about 5 minutes.



Figuur 1: Final design left side view

Figuur 2: Final design right side view

Now the mirror was ready to introduce to Mrs. M. She experienced, looking into the mirror, two normal upper limbs as is the case in other mirror therapy practices in patients with a more distal amputation. The mirror was de-assembled, carried by car to her home where she started her mirror box therapy. While in this patient the mirror therapy did not lead to a significant pain reduction, the design of the mirror was successful.

CONCLUSION

The new design succeeded in mirroring the amputee side, giving the patient the illusion of experiencing two normal upper limbs. With the present design, it is possible to perform mirror therapy in this group of forequarter amputee patients.

LONG TERM RESULTS OF EARLY MYOELECTRIC FITTINGS

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[2]*Centre for Rehabilitation Research, Box 1613, Sweden*

INTRODUCTION

There is an ongoing debate concerning the optimal age for first-time fitting of myoelectric prostheses to children. Sörbye advocates 2 ½ - 4 years of age as the best time [1] whereas centers in North America recommend fitting as early as 10-15 months of age [2]. The rationale for the early fittings is that these children will be more able to use the hand. But, at what age can children learn to operate a myoelectric hand? Which age for myoelectric fitting is the best considering the outcome in both short- and long time?

The aims of this study were to compare the age for i) voluntary operation of a myoelectric hand, ii) unrestricted operation of a myoelectric hand, and, iii) to compare the use of prostheses at different ages in children fitted before 2 and after 2 ½ years of age.

Table 1: Sample demographics and age (in months) at fitting of first myoelectric prosthetic hand

	Gender	Laterality	Prostheses side	Level of deficiency*	First myo
Case 1/ Control 1	Girl/Boy	Bilateral	Right/Right	BE lower/ CA tot	18/36
Case 2/ Control 2	Boy/Boy	Bilateral	Right/Right	CA part/ CA tot	17/37
Case 3/ Control 3	Boy/Boy	Unilateral	Right/Right	BE mid/ BE mid	13/37
Case 4/ Control 4	Girl/Girl	Unilateral	Right/Right	BE upper/ BE upper	17/36
Case 5/ Control 5	Girl/Girl	Unilateral	Right/Right	BE mid/ BE mid	8/41
Case 6/ Control 6	Boy/Boy	Unilateral	Left/Left	BE upper/ BE upper	15/38
Case 7/ Control 7	Boy/Boy	Unilateral	Left/Left	BE upper/ BE upper	19/33
Case 8/ Control 8	Girl/Girl	Unilateral	Left/Left	BE mid/ BE mid	11/32
Case 9/ Control 9	Boy/Boy	Unilateral	Left/Left	BE mid/ BE mid	21/36

*BE=below the elbow; upper=upper third; mid=middle third; lower=lower third; CA=carpal; part=partial absence of carpals; tot=total absence of carpal bones

METHODS

A prospective longitudinal case-control design was chosen for the study. The data-collection started in June 1995 and ended when the last child in the study group had reached the age of 12 (March 2011). Before initiation of the study, informed consent was obtained from the parents, and, because of the possible increase in costs related to the early fitting in the study group, from their health care provider. For the control group, informed consent was obtained from the parents.

Nine children were selected for early fittings and nine children were matched to the study-group with regards to gender, side and level of deficiency (Table 1). Cases were born between January 1994 and March 1999. Inclusion criteria were:

- transversal reduction deficiency below the elbow
- living in, or in the vicinity of the fitting centre
- younger than 2 years at time of fitting
- passive prostheses at 6 months of age
- family structure stabile and parents used to prosthetics

Instrumentation

Main outcome variable was the *Skills Index Ranking Scale* (SIRS), an observational based method used to categorize a persons ability to operate a myoelectric hand [3]. The scale ranges from 1 to 14 where each step describes an increasing ability to operate the hand (Figure 1). Based on the observations of the child during play or performance of other daily tasks, the occupational therapist decides at which level the child performs. The validity of the SIRS has been tested and the order of the steps confirmed by Rasch-analysis (unpublished data 1997).

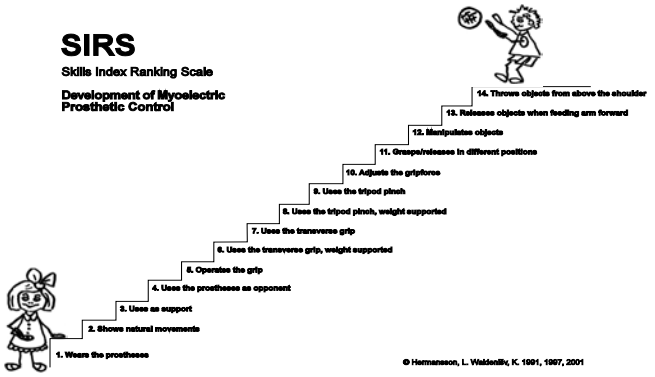

Figure 1: Skills Index Ranking Scale

The *Prosthetic Use Scale* (PUS) is a method for classification of the wearing pattern of a prosthesis. It is one of the outcome variables used in the Swedish quality register for upper limb prosthetics and amputation. The PUS is based on both wearing time and wearing pattern and ranges from 1 to 5 (Table 2). On the basis of the parents' or the child's reports, the occupational therapist scores the child on the scale.

Table 2: The Prosthetic Use Scale

1. *Full time*: uses the prosthesis more than 8 hours, 7 days a week.
2. *Part time*: uses the prosthesis 4-8 hours, 5-7 days a week.
3. *Occasional*: uses the prosthesis less than 4 hours, 1-7 days a week. The prosthesis is regularly used for one or more specific situations or tasks at least once a week, e.g., for use training or for meals.
4. *Sporadic*: uses the prosthesis at least once a month but not every week. The prosthesis is used sporadically some time every month, randomly.
5. *Non user*: has not used the prosthesis during the last month, or less.

Procedure

During a regular visit for control of prosthetic fit and use, a casting for a new socket was made. Three weeks later, fitting of the myoelectric hand took place at the clinic. In the study group the first myoelectric hand was fitted between 8 – 21 months of age (mean age 15.7 months, Table 1). These children were all fitted with an Otto Bock 2000 hand size 5", with a dual-site control system. Two children had parental access switch.

The children in the control group followed the regular fitting scheme and were, hence, fitted with a myoelectric hand at 32-41 months of age (mean age 36.2 months, Table 1). These children were fitted with an Otto Bock 2000 hand size 5 ½ " or 6", depending on the size of the contra-lateral hand, using a dual-site control system. No children had parental access switch.

Information to the parents about maintenance and use was the same in both groups. For the children in the study group, no formal training was initiated. Instead, parents were instructed to give verbal support and place objects in the hand, once they noticed action in the prosthetic hand.

After the children had reached three years of age, both groups of children were subject to regular training and support from the local team and/or the prosthetic clinic.

Follow-ups of the study group were initially made every third month. Between the age of 3 and 6, follow-ups were scheduled every 6th month, and after the age of 6, these were flexible and based on concomitant need for service or training. During the follow-ups the SIRS was performed and the parents were interviewed about the prosthetic use. Data from the control group were collected during regular follow-ups every 6- 12th month. The data were recorded in the patient files.

ANALYSIS

Wilcoxon Signed Ranks tests were performed to test for between-group differences. P-values lower than 0.05 were accepted as statistically significant.

RESULTS

Two subjects, one in each group, were lost for follow-up at 7 and 12 years of age, respectively.

Cases demonstrated voluntary control (SIRS 5 or higher) at 18 to 33 (median 24, inter-quartile range 21- 33) months of age, whereas in controls the corresponding age for this was 33 to 45 (median 36, inter-quartile range 34.5 – 40.5) months of age. The difference was statistically significant (p=0.015). One case showed voluntary control at time of fitting whereas all but one subject in the control group demonstrated voluntary control at time of fitting.

At 42 months of age, when all controls had been fitted, the median SIRS level was 7 in both groups (inter-quartile range: cases 5-10; controls 5-9; p= 0.674).

The highest ability with the myoelectric hand (SIRS 14) was first shown in the control group, at 5 to 9 (median 6, inter-quartile range 5.25 – 7.50) years of age, whereas in the cases this was shown more than one year later, at 6 to 12 (median 8, inter-quartile range 6 – 9) years of age. The difference was, however, not statistically significant (p=0.136).

The use of prostheses varied over the years and between the groups. However, although the median value was somewhat different, there were no statistically significant differences between the groups (Table 3).

Table 3: Prosthetic use at different ages (median; inter-quartile range)

Age (years)	Cases (n=8-9) Prosthetic use*	Controls (n=8-9) Prosthetic use*	p
3 ½	1.00; 1.00-1.50	1.00; 1.00-1.00	0.157
6	1.00; 1.00-2.50	1.00; 1.00-3.00	0.579
9	2.50; 1.00-3.75	1.00; 1.00-2.00	0.146
12	2.00; 1.00-4.75	1.00; 1.00-3.50	0.450

***1**=*Full time;* **2**=*Part time;* **3**=*Occasional;* **4**=*Sporadic;* **5**=*Non user*

DISCUSSION

The results from this study show that children can learn to operate a myoelectric prosthetic hand as early as 24 to 36 months of age. Earlier fittings result in earlier ability to reach the first level of control, but children fitted at the average age of 36 months do faster progression than the earlier fittings, resulting in a catching up at 42 months of age. This catching up is partly explained by the fact that all but one of the children fitted between 32-41 months of age were immediately able to operate the hand! This strongly supports fittings at around 3 years of age.

One interesting finding in this study was that the age range for development of ability to operate the myoelectric hand varied between the two groups of children. When looking at the inter-quartile range, the difference is six months for voluntary control and 9 months for unrestricted control. In both cases, the early fittings have the larger span for reaching the developmental stages. There are several plausible explanations for this. Clinically, we have noticed that children who are fitted before they have passed the "terrible two's", the age when most children demonstrate a strong integrity and will, often use the attention for their arm and prostheses to demonstrate their will. Hence, they do not want to wear the prostheses and do not attempt to operate the hand in order to use it for any purpose. This may result in a delay in ability to operate and use the prosthetic hand and could be one reason for the large variation in age for development of control. Other factors that probably have an impact on development of control are family climate and the parents' influence on training, routines in everyday life etc, and also the child's personality and learning pattern.

A factor that may influence the development of control is wearing time/pattern. High wearing time is expected to indicate high operational skill. From our experience, besides the parents' influence, wearing time is depending on service and support. Hence, technical problems with the prostheses may have a negative impact on the development. In this study we found no statistical significant difference in wearing time between the groups. There was, however, a greater tendency towards decreasing use of prostheses in the study group. Further studies with larger samples are needed to confirm this.

The major difference between the two groups of children was, besides the age, the training. In our centre training for children with a myoelectric prosthetic hand is based on the normal development of children. Once they reach the age of 3, most children are ready to cooperate and play with others. This forms a good basis for the training. The children who were fitted early did not receive any formal training until they reached the same age as in regular fittings. Despite that, although with a large age span within the group, they managed to learn to operate the hand earlier than the children with regular fittings. However, when looking at the results from when the children reached the highest ability to operate the hand according to SIRS, there is a tendency towards that the early fittings reach this later than the regular fittings. This indicates that the major impact of training is that it helps the child develop higher skills with the prosthesis and use it for daily activities. To understand the significance of training, studies on this topic is recommended.

There are several negative aspects to consider with early myoelectric fittings. One is the extra load that this puts on the parents at a usually were demanding time. Early myoelectric prosthetic fitting leads to an increase of visits to the health care provider. Furthermore, by the early fitting, much attention is given to the development of control instead of to the child. The question is, if the resulting ability to voluntarily operate the hand is worth it? Do children who have been fitted early have better use of their hand than those who have been fitted at a somewhat older age?

The major limitation of this study is the sample size. The number of cases is limited and, thus, requires a long period for data-collection. By increasing the number of controls, the study will gain power.

CONCLUSION

In conclusion, the best age for fitting of myoelectric prosthetic hands in children is around 3 years, with further consideration taken to the individual psychosocial development. Studies over the benefit from early fittings in daily life are needed.

ACKNOWLEDGEMENTS

We are indebted to the families who participated in this study. We also thank Kristina Waldenlöv, reg. OT and Malin Jansson, reg. OT, both at the Limb Deficiency and Arm Prostheses Centre, Örebro University Hospital, Örebro, Sweden, for contributing to the data-collection.

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CONTINUOUS AND SIMULTANEOUS EMG-BASED NEURAL NETWORK CONTROL OF TRANSRADIAL PROSTHESES

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ABSTRACT

As the development of dexterous prosthetic hand and wrist units continues, there is a need for command interfaces that will enable a user to operate these multi-joint devices in a natural, coordinated manner. In this study, myoelectric signals and hand kinematics were recorded as three able-bodied subjects performed a variety of individuated movements and simulated functional tasks. Time-delayed artificial neural networks (TDANNs) were designed to simultaneously decode the movement trajectories for seven distal degrees of freedom (pronation-supination, wrist ulnar-radial deviation, wrist flexion-extension, thumb rotation, thumb abduction-adduction, finger MCP flexion-extension, and finger PIP flexion-extension). Performance was quantified by calculating the variance accounted for (VAF) and normalized root-mean-square error (NRMSE) between the decoded and actual movements. Accurate predictions were achieved (VAF: 0.57-0.80, NRMSE: 0.04-0.11), suggesting that it may be possible to provide an intuitive EMG-based scheme that provides continuous and simultaneous multi-joint control for individuals with below-elbow amputations.

INTRODUCTION

While upper extremity amputations can cause a great deal of functional impairment, electrically-powered prostheses have proven to be effective tools for performing many daily tasks. While there has been a great deal of recent development in the mechanical design of prosthetic arms [1], a highly articulated limb is of little use if its movements are not well coordinated.

A number of different approaches have been taken to map EMG signals to the desired movements. Discrete movement types are often identified using pattern recognition approaches such as linear discriminant analysis, fuzzy logic, and artificial neural networks. Rather than classifying discrete states, the continuous prediction of trajectories has the potential advantage of enabling coordinated and simultaneous control of multiple joints. Reddy and Gupta [2] showed a direct relationship between surface EMG signals and joint angles during isolated single finger movements. This principle has also been used [3,4] to predict continuous movement trajectories for individual finger and wrist joints. The goal of the current study was to further investigate methods for decoding of continuous finger and wrist movements from electromyographic (EMG) activity of muscles located in the forearm using time-delayed artificial neural networks and develop a method to reduce the noise present in these decoded movements.

METHODS

Subject Information

3 male able-bodied subjects between the ages of 25 and 27 took part in these experiments. No subjects had a known history of any neuromuscular disorders. All subjects gave informed consent to the procedures as approved by the MetroHealth Medical Center Institutional Review Board.

EXPERIMENTAL PROTOCOL

An Optotrak Certus Motion Capture System (Northern Digital Inc., Waterloo, Ontario) was used to record the three-dimensional motions of the arm. Additionally, a CyberGlove II (CyberGlove Systems LLC, San Jose, CA) was used to measure the kinematics of the hand, wrist, and fingers via resistive bend-sensors. Surface EMG signals were recorded from an array of eight equally spaced electrodes around the circumference of the forearm. As in [5], the markings for the electrode array were positioned at 40% of the distance from the medial epicondyle of the humerus to the styloid process of the ulna.

Trials were collected while the subjects performed a variety of movements. Isolated movements involved moving a single degree of freedom at a time (e.g. flexing and extending the wrist or the metacarpophalangeal joint of the index finger). While motion of an individual digit is frequently accompanied by mechanically-coupled movements of adjacent fingers, subjects were instructed to not oppose these movements. Coupled movements involved moving multiple joints in concert, such as flexing and extending the fingers together or forming palmar, lateral, and power grasps. In a separate set of tasks, the subjects were presented with a

number of objects of varying geometry arranged randomly on a lap height table. Subjects were instructed to conform their hand to the objects without exerting much grasp force. During all trials, kinematic and EMG data were simultaneously recorded.

Data Processing

The digitized EMG data sets were then processed offline by filtering, windowing, and extracting signal features. The data were first high pass filtered to remove movement artifacts. Several features were then extracted from 128 ms rectangular windows of these signals with 50% overlap between adjacent segments. The time domain statistics described by Hudgins, Parker, and Scott [6] were used, generating a four-element feature set for each EMG channel.

As a practical consideration, several of the digitized joint angles obtained from the CyberGlove II were excluded from subsequent analyses. A list of the sensors used is provided in Table 1. The locations of the various bony landmarks measured by the Optotrak system were processed to obtain the pronosupination joint angle. The motion analysis data (from both the CyberGlove II and the Optotrak system) was then re-sampled and binned using a 128 ms window with 50% overlap. The average joint angle values during each window were used so that the sample time of the motion analysis data matched that of the EMG features. All kinematics were normalized such that 0 to 1 represented the full range of motion of each respective joint.

Table 1. Movements predicted in this study and how they were recorded

Movement	Motion Capture Method
Pronosupination	Optotrak
Wrist Flexion-Extension	CyberGlove
Wrist Ulnar-Radial Deviation	CyberGlove
Thumb Adduction-Abduction	CyberGlove
Thumb Rotation	CyberGlove
Middle Finger MCP Flexion-Extension	CyberGlove
Middle Finger PIP Flexion-Extension	CyberGlove

Neural Network Training

We investigated the use of a time-delayed artificial neural network with 20 hidden layer neurons and 5 input time delays to predict hand and wrist joint angle trajectories based on EMG information obtained from muscles that should be intact and available for recording in transradial amputees. A two-layer feed forward structure with a nonlinear tangent-sigmoidal activation function for the hidden layer and a linear output layer was utilized. All TDANNs were trained using backpropagation as implemented in MATLAB's Neural Network Toolbox (The Mathworks Inc., Natick, MA).

The performance of the TDANN was quantified by the normalized root mean square error (NRMSE) and the variance accounted for (VAF) between the experimentally recorded joint angle trajectories and the corresponding trajectories predicted by the TDANN. A 5-fold cross validation was performed, and all results represent the average across the five folds.

An "Adaptive" Filter for Improving Decoder Robustness

To regulate the neural network predictions, we implemented an "adaptive" moving average filter for each joint. When the probability of movement intent is high, the filter speeds up (reduces the width of the window) to track predicted rapid changes in the joint angle trajectories. Alternatively, when the probability of movement intent is low, the filter slows down (increases the width of the window) to reduce noise and increase smoothness.

A Bayesian approach was adopted for estimation of the probability of movement onset and offset. We defined a two class problem for each joint in which the two classes are determined by whether or not the joint is "active" (i.e. the joint velocity exceeds a predetermined threshold). The posterior probability calculated from Bayes' theorem was then used to gate the number of samples averaged by the filter. In addition to the goodness of fit measures previously discussed (NRMSE and VAF), the smoothness of movements was quantified as the number of peaks in the velocity profile. Fewer peaks in speed represent fewer periods of acceleration and deceleration, making a smoother movement. In this study, the number of velocity peaks (NVP) was calculated by counting the number of local maxima in the velocity profile that were greater than their respective preceding local minimum by at least 10% of the maximum velocity across all trials.

RESULTS

Figure 1 shows the average (± standard deviation) cross-validated TDANN prediction performance for each of the joints considered in this study. Shown in white is the mean performance for the unadjusted predictions, while the performance after applying the filter is shown in gray. In most cases, t. here is no significant difference in either VAF or NRMSE when applying the filter. Figure 2 shows the average (± standard deviation) NVP for each joint. Again, white bars represent the mean performance for the unadjusted predictions, while gray bars represent the performance after

applying the filter. In most cases, the NVP after adjustment are significantly decreased, suggesting that the filter does indeed smooth the movement and reduce the noise present in the predicted trajectories.

CONCLUSIONS

We have used TDANNs to decode continuous movements of seven finger, thumb, and wrist joints based on features extracted from EMG signals. These preliminary results show that there is significant information in these signals related to these movements and that there is the potential for providing users with continuous and simultaneous control of most of these joints. We have also demonstrated that an estimate of the probability of intended movement can be used to vary the characteristics of a filter to make the decoded movements more robust. More subjects will be included in this study to validate the results. Additional evaluation with amputees will also be necessary to determine if the findings in individuals with intact limbs will translate to the control of transradial prostheses.

ACKNOWLEDGEMENTS

This work was supported in part by a grant from DoD / TATRC (W81XWH-07-2-0044) and by a training grant from the NIH/NIBIB (T32-EB04314). We would like to thank B. Memberg for his efforts on regulatory issues and the Functional Neural Interface Lab at Case Western Reserve University for use of the CyberGlove.

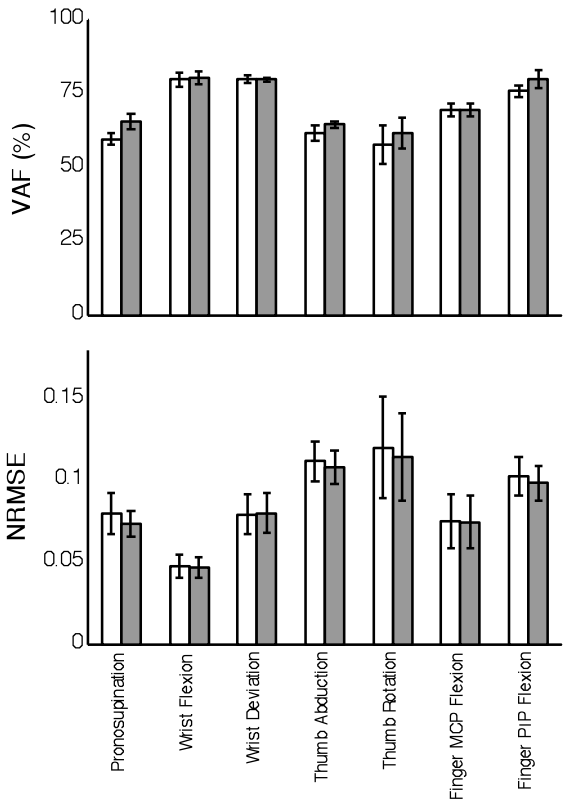

Figure 1. VAF (top) and NRMSE (bottom) of the predicted movement trajectories (mean ± SD). The unadjusted predictions are shown in white and the filtered predictions are shown in gray.

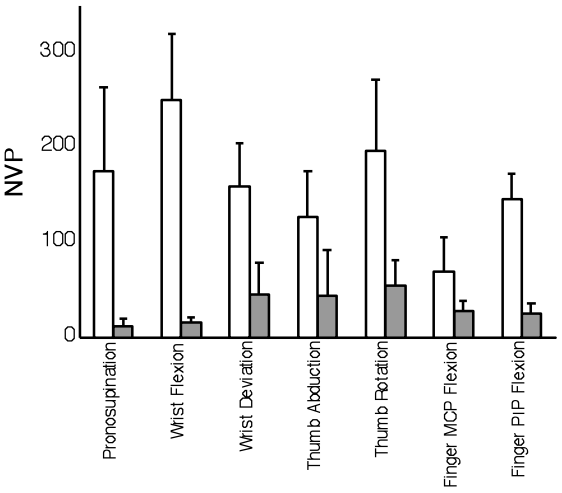

Figure 2. Numbe of velocitypeaks (NVP) in the predicted movements trajectories. The unadjusted predictions are shown in white and the filtered predictions are shown in gray.

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SENSOR OPTIONS FOR MULTI-ARTICULATING PARTIAL HAND PROSTHESES

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SUMMARY

The article gives an overview of various sensors options for partial hand prosthesis as a alternative to EMG sensors. The article analysed two kinds of Flexbend-Sensors and the Touch-Pad in combination with a electronic compensation of the battery-voltage for this sensors.

INTRODUCTION

Modern externally powered partial hand prostheses can now be equipped with up to 5 motorised fingers which allow separate movement of individual long fingers and thumb and their finger phalanges. Examples of this partial hand systems are the Vincent finger system [1][2][3] and the ProDigits [4].

The challenge is to provide safe and reliable operation of the high functionality for the amputees. The patient should have the choice of different grasp patterns which he may control proportional, see Figure 1.

Figure 1: cylinder grasp (l., m.) and index finger (r.), Partial hands with Vincent finger [J. Uellendahl, HANGER]

Especially for the control of the partial hand prostheses various control-options are required, as the requirement of each patient is very specific. Next to the most frequently applied EMG-Sensors, further input option for control of active prosthesis are available. In this context it may be reasonable to use the remaining mobility of fingers and the thumb for control purposes.

Two kinds of sensors are available for this purpose: Touch-pad and Flexbend-Sensors. The specific characteristics of such sensors are described here on the example of *FSR Sensor*, *Bend Sensor* and *Vincent-bend*, see Figure 2.

Figure 2: Sensors: *Vincent-bend*, *FSR Sensor* (Touch-Pad) and *Bend Sensor* in compare

ANALYSE OF FLEXBEND-SENSORS

The Touch-Pad as well as the both Flexbend-Sensors has a specific signal sequence, which should be considered for its successful application. In the subsequent section the sensors will be analysed and the characteristic of their signals will be described.

THEORY OF OPERATION FLEXBEND-SENSORS

A Flexbend-Sensor is a thin flexible sensor that changed in resistance when the sensor is bending. In the subsequent section two systems are introduced: the *Bend Sensor* from Flexpoint Inc. [5] and the *Vincent-bend* from Vincent Systems, a modification of the *Flex Sensor* produced by Abrams Gentile Entertainment Inc. (AGE) [6].

"Bend sensor consists of a single thin (.005"), flexible plastic film coated with a proprietary coating. This coating can also be used on other materials such as metals. When the sensor is bent, the coating separates into many micro cracks that open and close to the degree the sensor is bent. The opening and closing of cracks causes a measurable change in resistance."[5]

Flexbend-Sensor Var. (1): Bend Sensor (Flexpoint Inc.)

The *Bend Sensor Potentiometer* of Flexpoint Inc. is already available in a short mounting form of 25mm. Both

connecting cables are connected to one end of the sensor. The sensitive area of the sensor is approximately 10mm of longitude, see Figure 3.

Figure 3: *Bend Sensor* (Flexpoint Inc.) with wire

Flexband- Sensor Var. (2): Vincent-bend

The Flexband-Sensor *Vincent-bend* is a modification of the *Flex Sensor* from AGE Inc.. The sensor has been significantly shortened and equipped with new electrical contacts on both sides. The sensor is fittable for different lengths. The sensitive area of the applied version is app. 15mm long, see Figure 4.

Figure 4: *Vincent-bend* with wire

Metrological Analysis of Flexband-Sensors

For the purpose of the analysis the proximal end of the sensor has been fixated. The distal end has been bent along a measuring scale. The measurements have been taken in steps of 5°, see figure 4 for *Bend Sensor* and Figure 6 for *Vincent-bend*.

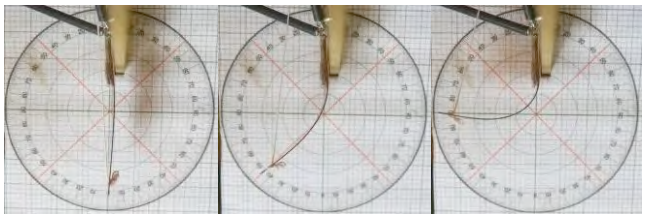

Figure 5: Measure *Bend Sensor* 0°, 45°, 90°

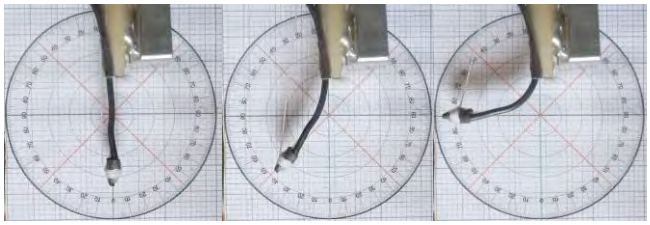

Figure 6: Measure *Vincent-bend* 0°, 45°, 90°

Hereby the resistance of the sensor has been measured. For the purpose of comparison of the sensors an sensor output voltage over a potential divider with a measuring resistance (Rmess) has been calculated for Bend-Sensor (Rmess 200kOhm) and for *Vincent-bend* (Rmess 22kOhm).

The comparison of the series of measurements shows different characteristics of the both Flexband-Sensors. The *Bend Sensor* of Flexpoint Inc. shows non linear characteristics. The sensitivity progresses with increasing bending. This affects the further process of the signal in the Controller. The modified *Vincent-bend* has an approximately linear progression and consequently an almost constant characteristic over the total measuring range. Its sensitivity is however slightly lower than the comparison specimen, see Figure 7.

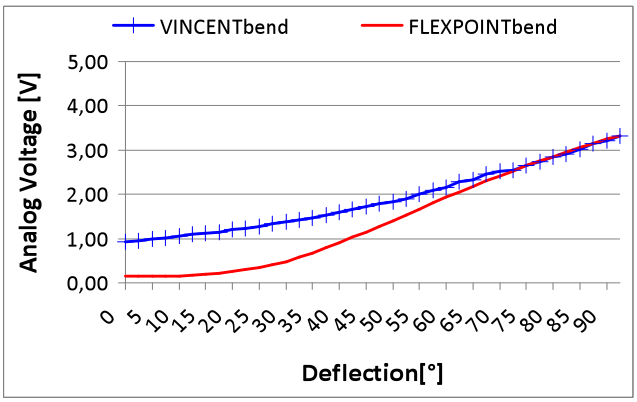

Figure 7: (Flexpoint) *Bend Sensor* (Rmess 200kOhm) and (AGE) *Vincent-bend* (Rmess 22kOhm)

TOUCH-PAD *FSR SENSOR*

The Touch-Pad is most a Force Sensing Resistor (FSR) Sensor from Interlink Electronics Inc. [7], a robust, polymer thick film (PTF) sensor devices, optimized for use in human touch control. Figure 8 show the *FSR Sensor* with wire.

Figure 8: Touch-Pad (*FSR Sensor*) with wire

Theory of Operation FSR Sensor

"The most basic FSR consists of two membranes separated by a thin air gap. The air gap is maintained by a spacer around the edges and by the rigidity of the two membranes. One of the membranes has two sets of interdigitated fingers that are electrically distinct, with each set connecting to one trace on a tail. The other membrane is coated with FSR ink. When pressed, the FSR ink shorts the

two traces together with a resistance that depends on applied force."[7]

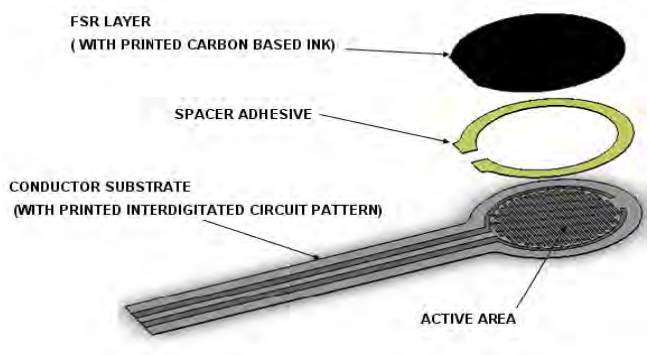

Figure 9: Basic *FSR Sensor* Construction [5]

Analyse of FSR Sensor

The signal sequence is relevant for the practical use of the *FSR Sensor*. With the following measurement setup the relationship of the pressure force to the resistance of the Sensor will be described. A cylinder is mounted in a squeezer, and an elastic half rounded cap is mounted to its tip. This cap presses in the center of a *FSR Sensor*, which lies on a highly sensitive scale. An increasing pressure is set up manually and the according resistance is measured, see Figure 10.

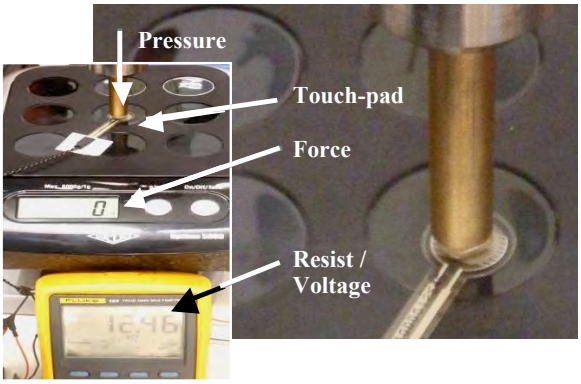

Figure 10: Measurement of the *FSR Sensor* characteristic

The field of characteristic lines has been converted by a measuring resistance (Rmess 22KOhm) of the FSR-resistor into an output voltage. The characteristic line of the *FSR Sensor* is non-linear. The sensitivity is high at low pressure and decreases with increasing force. Usually a *FSR Sensor* are applied at the analogue input of the controller instead of an EMG-electrode. For this purpose a resistor is placed between the analogue input and the mass to divide the potential.

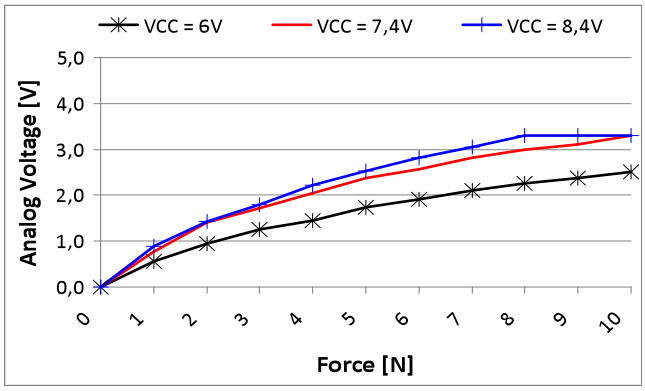

Figure 11: *FSR Sensor* (Rmess 22kOhm), 6-8.4Volt

At this set-up the battery voltage has a strong influence on the sensor signal. If it decreases, the signal level sinks as well, see Figure 11. The voltage range of a 2 cell Li-Pol accumulator lies between 8.4Volt in fully charged and 6.0Volt in discharged state. The output-signal of the *FSR Sensor* sinks hereby to approximately half of its original level. This effect can cause problems for a sensitive prosthesis control.

Software compensation of Battery voltage for FSR-Signal

The software on the Vincent controller measures permanently the variable battery-voltage and re-calculates the input sensor values, compensates the voltage drops, linearises the characteristic line of the sensor and filters artefacts.

Electronic compensation of Battery Voltage for FSR-Signal

In cases of absence of appropriate software, it may be solved electronically. For hardware compensation of the battery level, an electronic system was developed, the *Vincent-touch-s*. This stabilizes the reference voltage for resistance based sensors and makes them resistant against disturbances. At the same time it offers manual setting of the measuring resistance and this sensitive measuring range, see Figure 12.

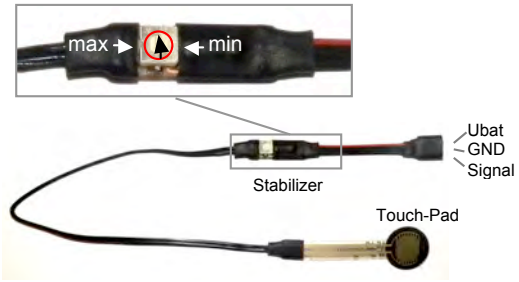

Figure 12: *VINCENT-touch-s* (s=stabilized)

The graphics shows the stabilized output signal for the three setting of the potential divider: min, middle and max.

At the same pressure on the sensor a higher output signal is obtained at the position „max. “, the sensor becomes more sensitive. See Figure 13.

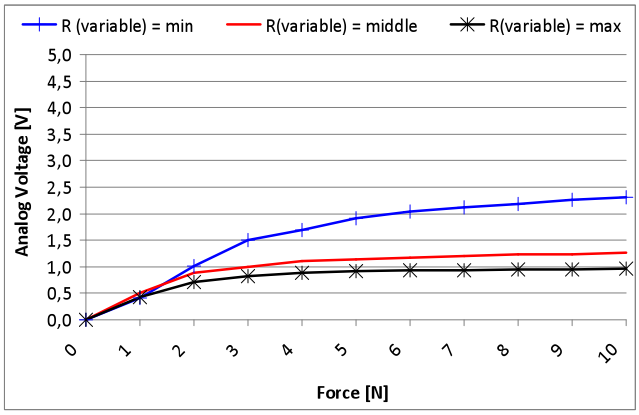

Figure 13: *Vincent-touch-s* (stabilized), min-max setting

FITTING WHIT FLEXBEND-SENSOR

As an example of a useful fitting with the control function of a Flexbend-Sensor, a two finger partial hand prosthesis of the company POHLIG will be described. The patient needs his left hand functional for his work as glassblower.

Patient situation and integration of the sensor

The index finger is completely maintained, but has a very limited ability to bend, half of the middle finger is ablated, and the ring finger and little finger are missing. The prostheses replace the missing two fingers with Vincent finger system. The index finger is used for the control of the prosthesis. In the area of its base joint a *Vincent-bend* Sensor is integrated in a silicon shank, see Figure 14.

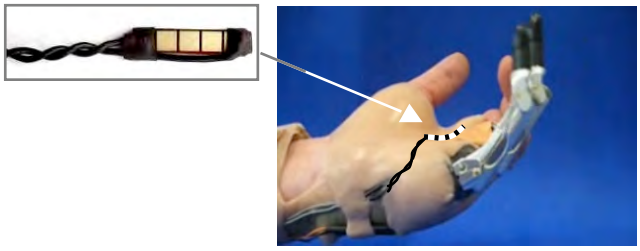

Figure 14: *Vincent-bend* sensor inside the silicon glove on position of index finger [M. Schaefer, POHLIG]

Control of the partial hand prostheses

The partial hand is controlled by the remaining function of the index finger. The principle of the control described in Figure 15 is a special mode of Vincent control software for use a *Vincent-bend* sensor.

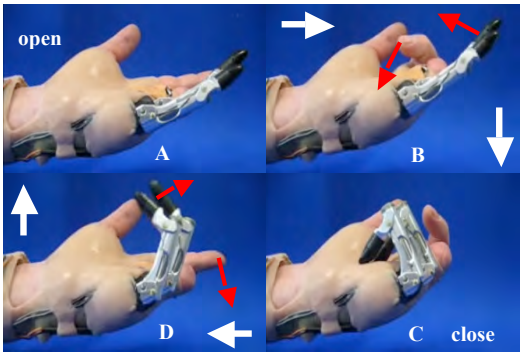

Figure 15: prostheses finger move from open-position (A) to close-position (C) and back to (A) during close (B) and open (D) of a single patient finger [M. Schaefer, POHLIG]

Bending the index-finger closes the fingers of the prostheses, straightening the index-finger the fingers open. Hereby the movements of opening and closing start only after exceeding an adjustable threshold. If the index finger stops, the fingers of the prosthesis stop as well. The faster the index finger and thus also the sensor are moved for purpose of control, the higher is the speed of the movement of the fingers of the prostheses.

Resume

First experiments showed that the use of a Flexbend-Sensor is helpful by the patients. The Sensor allows a very intuitive control of the partial hand prostheses. The patient can control the electrical finger after a short training period. The set-up is small and integrates well. However it is very new and the results of a longer clinical trail are yet to come.

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UNIQUE SITUATIONS IN PROSTHETIC DESIGN WHEN APPLYING TARGETED MUSCLE REINNERVATION IN TRANSHUMERAL AND SHOULDER DISARTICULATION LEVELS

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INTRODUCTION

After several cases utilizing TMR as the foundation for control it becomes apparent that even the most advanced commercially available systems lack all the necessary variability to accommodate a wide array of clinical presentations. Some patients present with multiple sites for control but may want, in addition to a fully myoelectric prosthesis, a hybrid device; giving them the flexibility to use all available types of prostheses. Some patients present with muscle that is hyper-mobile; making the job of securing the prosthesis and providing a good interface for the electrode an extreme challenge. As a result there is a need for further development of components made available so as to provide devices that meet these demands.

HYBRID TMR PROTHESIS WITH THREE SIMULTANEOUS DEGREES OF FREEDOM

Transhumeral hybrid prostheses have been used successfully by upper extremity amputees for many years to increase function while allowing 2 degrees of simultaneous freedom; controlling the elbow while operating the terminal device. Many patients prefer this prosthetic control strategy to the strict body powered or myoelectric designs. With the advent of recent TMR Targeted Muscle reinnervation surgical techniques, we have been able to increase function of hybrid prostheses.

Targeted Muscle Reinnervation has given Prosthetists additional myoelectric locations to capture EMG signals. With the increase in the number of EMG sites the Prosthetist can increase function for the patient. To date most patients who underwent TMR surgery were fit with purely myo electric system (myoelectric elbow, wrist and hand or other terminal device). In most cases this gives the user 2 degrees of simultaneous control of the myoelectric prosthesis.

Using existing hybrid transhumeral prosthetic components we are able to increase simultaneous control of the prosthesis from two to three degrees of simultaneous control. This allows the patient the ability to actively control the elbow flexion and extension, wrist pronation and supination, and hand opening and closing simultaneously giving the patient a more natural way of moving the prosthesis in space while doing activities of daily living more closely resembling a natural human arm.

Advantages of the three degrees of simultaneous movement should allow the patient faster response time when manipulating objects, greater wearing time due to the decreased weight of the body powered elbow and increased proprioception and speed of elbow flexion in space through the Bowden cable system.

Extensive Occupational and Physical therapy will need to take place to improve the patient's control and function with a device that allows the user three simultaneous degrees of freedom.

Overall, patients that have been fit with the Hybrid TMR prosthesis report they prefer to wear this prosthesis over their body powered or TMR myoelectric prosthesis.

Further trials will need to occur to in future TMR patients to gain the understanding of what can be accomplish.

THE SOCKET AND ELECTRODE INTERRELATIONSHIP IN TMR SHOULDER DISARTICULATION CASES

The goal behind TMR surgery is to surgically treat the nerve in a more appropriate manner, to produce more electrode sites and to create sites that are more intuitive resulting in less switching and higher simultaneous control of multiple degrees of freedom. Along with the creation of more electrode sites there are issues that may require the typical approaches to socket design and electrode placement and management to be modified. Sometimes the surgery produces a muscle that is hyper-mobile. This may be caused directly by hyper-reinnervation, surgically removing the origin or insertion of the muscle or by removing the sub-cutaneous tissue resulting in a more adherent skin to muscle connection. Regardless of the cause, the result is that managing the placement and constant contact between the electrode and the skin can be a challenging problem.

The problems of tissue sliding and pulling away from the socket have been previously reported.1 In Shoulder disarticulation cases this problem of hyper-mobility is more of an issue because frame type socket designs do not encapsulate the musculature and restrict movement as it would in a Transhumeral situation. This gives the skin over the muscle freedom to move and makes the job of keeping the electrode in place more of a challenge. If the electrode is held in position in the frame, we typically expect the skin to stay relatively in the same position under the electrode. If any sliding occurs, a motion artifact is produced resulting in unpredictable behavior of the prosthesis. It is common to allow for some flexibility outward with the electrode with flexible mounts but it has not been a problem to control lateral shifting of the musculature by simple socket designs. In these TMR cases, expansion, change in topography and lateral shift of the muscle is very common.

Initially efforts were made to modify the socket for the final resting position of the electrode upon complete contraction. When not contracting, the muscle and the surrounding soft tissue would find a home inside the strangely shaped interior of the frame. This worked well to a point but several side issues arose. The patients would report significant pressure over their electrode sites, they would describe numb feelings in their "transferred" limb sensation and often the muscle would not fire as strong resulting in the need for higher than needed amplification through the electronics. Another problem with having so many electrodes in one area, like over the pectoralis muscle, was that the socket surface had to be quite large to accommodate the electrodes. This made the socket larger, hotter and more susceptible to coming off of the chest when in a seated position from counter pressure to the posterior inferior member of the supporting socket frame. These issues spawned the thought that if we could develop an individual electrode holding appliance and connect it flexibly and remotely to the frame that we could control individually the tension over the site and independent of any other electrodes, keep them separately flexible when each muscle fired.

The first generation of flexible connector involved a spring steel arm and a fixed connector at the electrode that would pivot over the length of the spring steel. Otto Bock suction socket electrodes were utilized to provide the connection point to the fixed connector. If the only need was to accommodate substantial outward movement of the skin, more than a typical flexible electrode mount would accommodate this worked fine. It did not work however in a situation that the topography changed where the angle of the electrode needed to change to maintain contact. Tilting of more than 30 degrees was necessary in this case.

Another generation of device was made with a pivoting attachment over the electrode. It accommodated the necessary tilt of the electrode but lacked the ability to control rotation of the electrode. It was also noted that this design allowed for some accommodation of the shear movement under the electrode by compressing the soft tissue over the muscle as it expanded. This isn't a perfect solution when the skin is significantly sweaty as there is less coefficient of friction between the electrode and the skin so the tilt would not be enough to stabilize the electrode.

The third generation under development will accommodate the following criteria

1. Spring steel attachment arm for adjustable tension and flexible attaching to the socket frame
2 Rotatable locking electrode holder
3. Pivoting head
4. Gain adjustment access
5. Shear accommodation
6. Protection for electrode wire

With the above listed modifications the device will be able to be used on even the most challenging of TMR presentations and maintain independence of the electrode from the anchor and necessarily stable structure of the socket.

CONCLUSION

TMR surgery has expanded the functional capacity of modern upper limb prosthetic devices and as we explore these cases and try to maximize the potential of each patient it is clear that there is a great need for fine tuning our approaches. As a result there will be a spill-over effect of these techniques incorporated for both the TMR population and non-TMR population to benefit..

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METHODS FOR COLLECTING MYOELECTRIC SIGNALS FROM INDIVIDUALS WITH LOWER LIMB AMPUTATIONS

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ABSTRACT

Technological advancements in lower limb prostheses have resulted in actuated motors in both knees and ankles. Currently, these components are controlled by information measured from various electromechanical sensors attached to the prosthesis. Our aim is to enhance the control information provided to powered prosthetic components by including input from the user via interpreted myoelectric signals (MESs). To extract useful control information, it is imperative that consistent, high-quality MESs be collected from patients each time they don the socket. In this work, we present approaches to maintaining consistent electrode placements on individuals with transfemoral and transtibial amputations during static non-weight-bearing conditions and dynamic weight-bearing activities. Our results show that a variety of methods, similar to those used in upper limb fittings, may be used to collect high-quality MESs during static non-weight-bearing conditions. MES collection during dynamic weight-bearing activities is more challenging. The type, size, shape, and placement of electrodes must be carefully chosen to maintain contact with the skin without compromising comfort during weight-bearing activity.

INTRODUCTION

There are an estimated three million individuals in North America with major amputations [1], with an estimated 90% to 97% being lower limb amputations [2]. Most lower limb prosthetic components are passive, reacting to the external forces applied to them. Powered lower limb components consist primarily of microprocessor knees, which use input from electromechanical sensors to alter the resistance of the knee unit to compensate for different phases of the gait cycle or variations in cadence. Until recently, the only components that contained a motor-actuated joint for positioning were the Ossur Power Knee™ and Proprio Foot™ and the Power Knee was the only commercially available prosthetic component that actually generated positive power, which may reduce the user's energy expenditure and improve gait mechanics [3].

Powered lower limb components with actuated motors have been developed and tested clinically and are highly visible in the research community, with the PowerFoot BiOM™ by iWalk recently becoming commercially available. Each mode of operation of these components (e.g. stair ascent) has a kinematic profile that determines the operations of the joint. Although highly sophisticated, this variety of powered component still relies on electromechanical sensors to trigger a particular mode. Switching between modes can also be done manually. Such methods for control are not intuitive, do not provide smooth transitioning between modes, and can be cumbersome as they may involve use of the contra-lateral limb and may require donning additional hardware.

In order to enhance the performance of these lower limb prostheses, it is our goal to augment the current sensor information with user intent information. Our approach to this merger of technology is to use MESs from the surface of the individual's residual limb to provide data that will improve component responsiveness. A study by Huang et al. [4] investigated the use of advanced signal processing as a control strategy for powered lower limb prostheses. The results indicated that the combination of surface MESs and pattern recognition can provide accurate information regarding the user's intent for prosthetic control. In order to utilize user intent information, it is essential to create an interface that captures consistent, reliable, high-quality MESs from residual limb muscles during both static and dynamic situations [4].

There are standard practices regarding the incorporation of electrodes into upper extremity prostheses. Two main methods are identified; first being the *packaged electrode*, which is a combination of contacts and pre-amplifiers. These are typically rectangular-shaped packages that are mounted to the inner socket (or interface) that is in direct contact with the skin. Appropriate placement of these packaged electrodes is crucial when fabricating the interface, as these cannot be readily repositioned without creating a void in the socket and remounting the package. *Remote electrodes* are those in which the electrode contacts are separate from the pre-

amplifier. A pair of contacts and a single reference is usually associated with each amplifier. These are convex or dome-shaped medical-grade stainless steel and come in different diameters and heights. Daly [5] described the use of remote electrodes with gel liners in 12 upper limb subjects. Although he reported an improvement in comfort and function, durability of the electrodes and wiring still pose challenges in this design [5]. Advantages to using remote electrodes are that the contacts can be easily placed at different locations within the socket/interface, they can be spaced apart from one another at varying positions and moved to another location easily without creating a large void, they can be placed at varying in depths relative to one another, and they can be mounted in irregular contours (convex or concave aspects of the socket).

METHODS

All research activities were approved by the Northwestern University Institutional Review Board.

Transfemoral Fittings

The advantages of remote electrodes and the amount/type of soft tissue present in most transfemoral limbs provided an ideal combination for collection of surface MESs with transfemoral sockets. However, early in our research the results using remote electrodes were suboptimal. In the study by Huang et al. [4], dome-style contacts (Liberating Technologies, Inc.) were incorporated into a transfemoral diagnostic suction socket by drilling holes precisely 18 mm apart—the spacing of the MA-411-002 electrode (Motion Lab System, Inc.)—as the electrodes were mounted directly onto the threads of the contacts [4]. The threads of the contacts had to be parallel to one another and spaced at a distance to allow them to screw into the socket-mounted electrodes. If this constraint was not precisely met, the holes had to be re-countered. This resulted in oversized holes that compromised suction. We attempted to remedy this by applying silicone putty between the contact and electrode in order to re-establish suction. This was found to be time-consuming and tedious.

Later, we established a different protocol for collecting MESs from subjects with transfemoral amputations [6]. MESs were first collected during a static, non-weight-bearing condition without a prosthesis or socket. Nine muscles were identified on the residual limb, including sartorius, rectus femoris, vastus lateralis, vastus medialis, gracilis, adductor magnus, semitendinosus, biceps femoris, and tensor fascia latae. Self-adhesive Ag/AgCl contacts were applied over these sites and were snapped to modified surface MES sensors (DelSys). In an attempt to keep electrode locations relatively consistent between static and dynamic conditions, the positions of these electrodes needed to be re-located onto a test socket. If a well-fitting diagnostic socket had been previously fabricated, the subject was asked to don the socket multiple times and the muscle sites were marked on the socket. The average muscle location during these donning attempts was then used to locate the socket-mounted contacts. If a diagnostic socket was not available, an impression was taken with fibreglass bandage and the electrode locations were later transferred to the test socket.

With the DelSys electrodes, it was possible to use a different style of stainless steel dome contact (Motion Control, Inc.) within the socket. These contacts permitted a snap, analogous to those on the self-adhesive Ag/AgCl

Figure 1: Transfemoral test socket with domes and snaps mounted for MES collection.

contacts, to be mounted on the outside of the socket. Domes were then threaded through the diagnostic socket and into the back of the snap (Figure 1). This greatly decreased the time and complexity of the diagnostic socket set-up. Data with the socket could then be collected in either static or dynamic conditions.

Transtibial Fittings

Transtibial sockets present a different challenge due to anatomical contours and minimal soft tissue coverage. Typically, a soft interface (i.e. sock and/or liner) exists between the skin and hard socket to provide comfort and/or a means of suspension. Our team did not feel it was plausible to attempt to fit individuals with transtibial sockets that were similar to the transfemoral designs, as the residual limb would need to be in direct contact with the hard socket and stainless steel domes, compromising comfort and electrode-skin contact. Two alternative approaches were (1) to place contacts on the skin prior to donning the soft interface and socket, or (2) to embed contacts into the soft interface itself. We chose to examine the latter, as many individuals with transtibial amputations utilize gel liners and this approach is in line with our group's ongoing research into gel liners.

The method that we chose to employ was different from that described by Salam [7], who cut holes in the liner for residual limb/electrode contact, or Daly [5], who used snap electrodes through the interface along with a pre-amplifier wire harness. Incorporating DelSys electrodes permitted us to design a liner that would contact the residual limb at the required muscle locations and then carry this information to a remote location where snaps could be used to connect to the DelSys electrodes. As in our previous designs, requirements for the interface were that it (1) was easily donned and doffed,

(2) was comfortable, and (3) contained flexible leads to permit bending and rolling without fear of fatigue or damage.

Our first transtibial subject had been using an Iceross Synergy Liner™ by Ossur and was therefore accustomed to donning and doffing techniques. We modified this liner for MES collection at eight desired electrode sites: over the rectus femoris, biceps femoris, vastus medialis, vastus lateralis, gastrocnemius (lateral head), gastrocnemius (medial head),

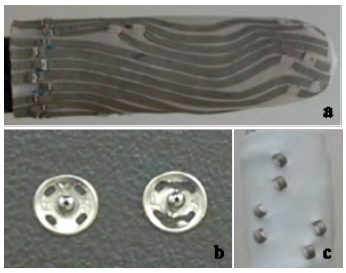

Figure 2: a) Modified Liner; b) Snaps; c) Inner Contacts with Domes

tibialis anterior, and the peroneus longus. These muscles were palpated and marked, and marks were transferred onto the roll-on gel liner. Two contacts were required for each bipolar electrode, and slits were cut in the liner to weave conductive fabric through to the inside and back out. Each contact site was 1 cm wide and 2 cm long with a 3 cm center-to-center distance between contact sites for each bipolar pair. Contacts were made on a slight angle to allow the fabric to travel up and down the liner without touching adjacent fabric strips. The conductive fabric was secured to the outside of the liner, terminating in snaps (Figure 2).

On the subject's first visit, we identified muscle sites and performed non-weight-bearing MES testing. Self-adhesive Ag/AgCl contacts were used for this experiment. The subject was familiarized with the protocol and MESs were collected while he visualized performing different movements with his missing limb in a static, non-weight-bearing condition.

The subject came in on three additional days to test MES collection with the modified liner. At each visit, the subject was asked to perform muscle contractions for the same motions introduced on the first day. MESs were collected under three conditions: (1) with the liner and no socket, (2) with the liner and socket but non-weight- bearing, and (3) with liner and socket during walking trials. At the first visit, the liner was tested without anything under the fabric to raise the contact sites (the fabric was flush with the gel liner). At the second visit, small leather discs were glued under the fabric inside the liner to raise the contact site from the surface of the gel liner in order to improve contact with the subject's residual limb. For the third visit, we used higher silicone domes instead of the leather discs; again the goal was to achieve and maintain good contact with the subject's limb without compromising comfort.

RESULTS

Transfemoral Fittings

The results of data collection have shown promise for the new socket design both statically, in a seated position, and dynamically, with both a passive and powered prosthesis.

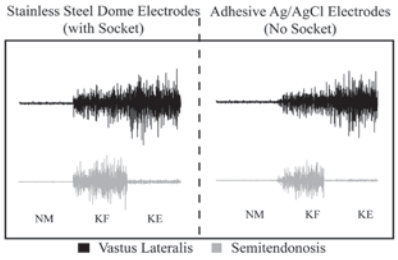

Figure 3: MESs from a transfemoral amputee during non-weight-bearing activities: no motion (NM), knee flexion (KF), and knee extension (KE).

Only minor modifications, typical in prosthetic fittings, were necessary to the socket/contact interface. There were minimal differences between the myoelectric signals recorded from the Ag/AgCl electrodes without a socket and the stainless steel dome electrodes embedded into the socket (Figure 3). These data were recorded on separate days so small differences in signal amplitude can be attributed to differences in electrode position (i.e. donning the socket) and muscle contraction intensity.

Each electrode setup was used to train a pattern recognition system for both knee and ankle motions in the sagittal plane. The system was 93% accurate using the Ag/AgCl electrodes without a socket and 92% accurate using the stainless steel dome electrodes embedded into the socket.

During weight-bearing activities, increasing the depth of the contacts decreased motion artefact and potential lift-off within the socket. Lift-off is most often characterized by large signal amplitudes with a 60 Hz frequency component and usually occurs during heel strike and/or toe off. The addition of spacers behind the convex dome or aggressive modification of the positive model and/or diagnostic socket has reduced the likelihood of lift-off. Using these modifications, we were able to use stainless steel dome electrodes to collect high-quality myoelectric signals during walking (Figure 4).

TRANSTIBIAL FITTINGS

Transtibial data collection has also proven comparable to other methods of obtaining myoelectric signals during non-weight-bearing conditions. When compared to signals obtained using Ag/AgCl contacts, the myoelectric signals displayed from the medial gastrocnemius and tibialis anterior

show similar characteristics (Figure 5). When used to train a pattern recognition system for ankle motions in the sagittal plane, the system was 100% accurate using the Ag/AgCl electrodes and 100% accurate using the fabric electrodes with a socket.

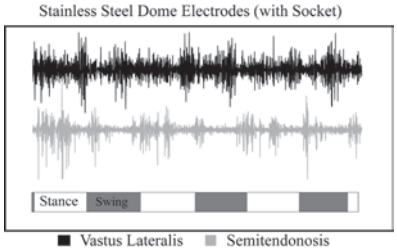

Figure 4. MESs from a transfemoral amputee while walking.

In dynamic weight-bearing conditions, myoelectric signals were not as clean, with a movement artifact present during peak periods of loading and unloading (Figure 6).

DISCUSSION

For our current research with transfemoral subjects, it is plausible to use a test socket with dome-style electrodes and snaps. However, in future developments, it may be necessary to alter the configuration to permit the inner socket and/or frame to contain the wire harness, or to use a liner in conjunction with transfemoral fittings and house the electronics somewhere within the prosthesis itself. However, this may compromise the fit and control of the prosthesis. We feel that fitting liners to individuals with transfemoral amputations is less optimal than fitting traditional suction sockets.

Within the transtibial MES recordings it is difficult to surmise exactly what is occurring inside the socket and

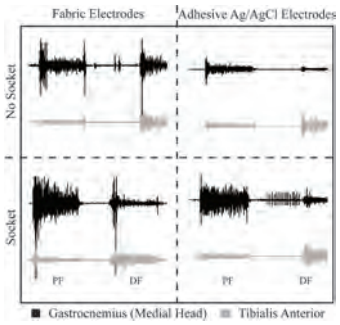

Figure 5. MES from a transtibial amputee during a non-weight-bearing session performing ankle plantar flexion (PF) and dorsiflexion (DF) muscle contractions.

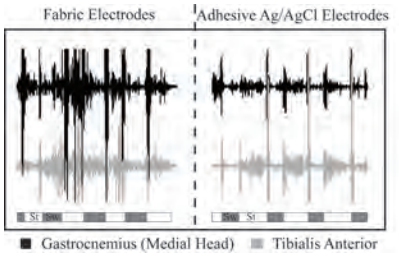

Figure 6. Myoelectric signals from a transtibial amputee while walking. St = stance; Sw = swing.

interface, although the data appears to suggest a lift-off of one or more of the contacts from the skin. Pistoning (translational movement of the limb within the socket) or movement of subcutaneous tissue may also be the cause of such artefacts. Deepening the contacts on the muscle bellies proved effective in the collection of MES, however, this was done at the expense of comfort. New style contacts are being investigated to improve the reliability of the signals as well as the comfort for the user.

ACKLOWEDGEMENTS

This work was supported by the US Army Telemedicine and Advanced Technology Research Center (TATRC) project W81XWH-09-2-0020.

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REAL-TIME RECOGNITION OF USER INTENT FOR NEURAL CONTROL OF ARTIFICIAL LEGS

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INTRODUCTION

Lower limb amputation significantly affects the quality of the leg amputee's daily life. Recent advancements in embedded electronics and electromechanical actuators have propelled the recent development of powered artificial legs [1-3]. Usually, finite-state machine (FSM) is utilized in the design of powered prosthetic legs to control the knee joint impedance or knee position in each gait phase [2, 4]. The impedance adjustment of the powered knee depends on the locomotion modes [2-3], since the dynamics and kinematics of the knee joint varies across different locomotion modes. Thus, in order to allow the prosthetic leg appropriately select the prosthetic control mode and smoothly transit the activities from one to another in time, the user must "tell" the prosthetic leg the locomotion intent before execution of the transitions. Currently, the artificial legs are manually controlled by using exaggerated hip and trunk motion [4], which is cumbersome and sometimes unreliable. Accurately recognizing the leg amputee's locomotion intent is required in order to realize the smooth and seamless control of prosthetic legs.

An intent recognition approach for the real-time control of a powered lower limb prosthesis, which utilized the mechanical sensor information, has been reported in a recent study [5]. One patient with transfemoral (TF) amputation performing level-ground walking, sitting, and standing was tested. The study reported 100% accuracy of recognizing the mode transitions and only 3 misclassifications during a 570s testing period. However, over 500ms system delay was reported, which may be inadequate for users to perform safe and smooth locomotion transitions. In addition, gait initiations and terminations were the only locomotion transitions tested. Only using mechanical information may not be able to promptly recognize the transitions between different locomotion modes because this type of information may not necessarily correspond with the user's intent. Alternatively, utilizing the neural control signal may enable the true intuitive control of the artificial limbs.

As one of the major neural control sources for the powered prosthesis, surface electromyographic (EMG) signals have been successfully applied in the control of upper limb prosthesis [6-9]. However, the EMG pattern recognition methods used in upper limb control cannot be directly applied on the lower limb prostheses, due to the non-stationary characteristic of EMG signals measured from the lower limb muscles during dynamic locomotion movement. In order to address this challenge, a phase-dependent EMG pattern recognition strategy was developed in our previous study [10]. This approach was tested on eight able-bodied subjects and two subjects with TF amputation. About 90% accuracy was obtained when recognizing seven locomotion modes. In addition, the user intent recognition accuracy was further improved by a neuromuscular-mechanical fusion algorithm [11], which fused EMG signals measured from the residual thigh muscles and the ground reaction forces/moments collected from the prosthetic pylon. The algorithm was tested in real-time to recognize three locomotion modes (level walking, stair ascent, and stair descent) on one able-bodied subject with 99.73% accuracy.

Although the experiment on the able-bodied subject has demonstrated promising results, whether or not the designed intent recognition system can be used for neural control of artificial legs is unclear. This is because there might not be enough EMG recording sites available for neuromuscular information extraction due to the muscle loss in patients with leg amputations, which may cause the accuracy of user intent recognition to be inadequate for robust prosthetics control. Therefore, in order to evaluate the potential of the intent recognition system for prosthetic legs, the designed system was evaluated on one TF amputee subject via real-time testing. In addition, besides the previous tested tasks, another two tasks: sitting and standing, were also included in this study. It is hoped that the results of this study could aid the further development of neural-controlled artificial legs.

METHODS

Structure of User Intent Recognition System

The whole structure of the intent recognition system is demonstrated in Fig.1. The multichannel EMG signals and mechanical measurements are simultaneously streamed into the system and then segmented into continuous, overlapped analysis windows. EMG features from each channel and the

mechanical features from individual degree of freedom were extracted in each analysis window and further concatenated into one feature vector. The fused feature vector is then sent into a phase-dependent classifier. The phase-dependent classifier consists of multiple sub-classifiers, each one of which is established based on the data in one defined gait phase. The gait phase detector detects the current gait phase and switches on the corresponding sub-classifier. A post-processing algorithm is applied to the decision stream to produce smoothed decision continuously.

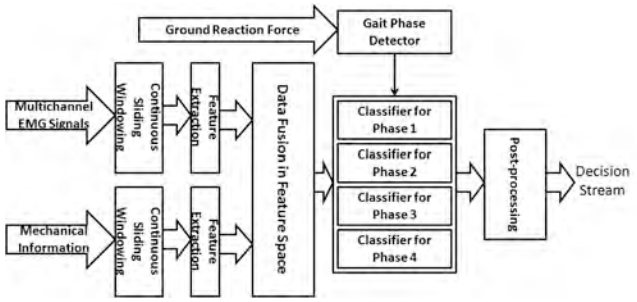

Fig. 1. Structure of intent recognition system based on neuromuscular-mechanical fusion.

Sensor Data Pre-processing and Feature Extraction

An eighth-order band-pass Butterworth filter with cut-off frequency between 25 and 450 Hz is applied on the raw EMG signals. The mechanical forces/moments recorded from the load cell mounted on the prosthetic pylon are low-pass filtered with a 50 HZ cut-off frequency. Then, the signal streams are segmented by sliding analysis windows as shown in Fig. 2. In this study, the length of the analysis window is 150 ms and the window increment is 50 ms.

Four time-domain (TD) features were extracted from the EMG signals: (1) the mean absolute value, (2) number of zero crossings, (3) number of slope sign changes, and (4) waveform length as described in [8]. For mechanical signals, the mean, minimum, and maximum values in each analysis window were extracted as the features.

Phase-dependent Classification Strategy

Different from the discrete gait phases with constant 200ms duration proposed in our previous study [10], continuous gait phases were used in this study. Four clinical gait phases are defined (shown in Fig. 2). The real-time gait phase detection is implemented by monitoring the vertical ground reaction force (GRF) measured from the load cell mounted on the prosthetic leg. The detection criteria are shown in Fig. 3. The applied contact threshold is 2% of the subject's weight. If one analysis window is located between two defined gait phases (e.g. the window W2 Fig. 2), the activated classifier is associated with the gait phase, in which it incorporates the data more than half of the window length (e.g. the classifier associated with the phase 2 should be used for the data in W2).

A Support Vector Machine (SVM) classifier with a nonlinear kernel is used in this study. A multiclass SVM with "one-against-one" (OAO) scheme [12-13] and C-Support Vectors Classification (C-SVC) [14] are used to identify different locomotion modes. The applied kernel function is the radial basis function (RBF). A 5-point majority vote scheme is applied to eliminate the erroneous decisions from the classifier. More detailed information about SVM algorithm can be found in [13-14].

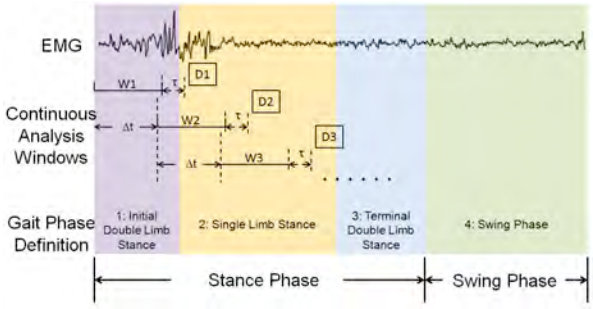

Fig. 2. Continuous windowing scheme for real time pattern recognition and definition of gait phases. For each analysis window (W1, W2, and W3), a classification decision (D1, D2 and D3) is made Δt seconds later. τ is the processing time required of the classifier, where τ is no larger than Δt.

Fig. 3. The real-time gait phase detection criteria

Subject and Experimental Setup

This study was conducted with Institutional Review Board (IRB) approval and informed consent of the subject. One female patient with unilateral transfemoral (TF) amputation was recruited. Eight channels surface EMG signals from the residual thigh muscles were collected by an EMG system (Motion Lab System, US) and used for intent recognition. The EMG electrodes were embedded in customized gel liners (Ohio Willow Wood, US) for both comfort and reliable electrode-skin contact and placed at locations where strong EMG signals could be recorded. A ground electrode was placed on the bony area near the anterior iliac spine. The EMG system filtered signals between 20 Hz

and 450 Hz with a pass-band gain of 1000 and then sampled at 1000 Hz. Mechanical ground reaction forces and moments were measured by a six–degree of freedom (DOF) load cell (Bertec Corporation, OH, US) mounted on the prosthetic pylon. The forces/moments were also sampled at 1000 Hz. All data recordings were synchronized and streamed into a PC through data acquisition system. The real-time algorithm was implemented in MATLAB and the real-time locomotion predictions were displayed on a flat Plasma TV. In addition, the states of sitting and standing were indicated by a pressure measuring mat which was attached to the gluteal region of the subject.

Experimental Protocol

The subject wore a hydraulic passive knee during the experiment period. Experimental sockets were duplicated from the subject's ischial containment socket with suction suspension. The subject received instructions and practiced the tasks several times prior to experiment.

Three locomotion modes including level-ground walking (W), stair ascent (SA), and stair descent (SD) and two tasks such as sitting (S) and standing (ST) were investigated in this study. The resultant mode transitions included W→SA, SA→W, W→SD, SD→W, S→ST, ST→W, W→ST, and ST→S. The experiment consisted of two sessions: training session and testing session. The training data collection for building the classifiers was performed in the training session. At least three training trials for each task were required in order to collect enough training data. During the real-time testing session, the subject was asked to continuously transit among the five different tasks. Each trial lasted around one minute. Totally 15 real-time testing trials were conducted. For the subject's safety, she was allowed to use hand railing. Rest periods were allowed between trials to avoid fatigue.

Real-time Performance Evaluation

The real time performance of intent recognition system is evaluated by the following parameters.

1) *Classification Accuracy (CA) in the Static States:* The static state is defined as the state of the subject continuously walking on the same type of terrain (level ground and stair) or performing the same task (sitting and standing). The classification accuracy in the static state is quantified by

$$CA = \frac{Number\,of\,correctly\,classified\,observations}{Total\,number\,of\,observations} \times 100\% \quad (1)$$

2) *The Number of Missed Mode Transitions:* For the transition between different locomotion modes, the transition period starts from the initial prosthetic heel contact (phase 1 in Fig. 2) before switching the negotiated terrain and terminates at the end of single stance phase (phase 2 in Fig. 2) after the terrain switching; for the transition between different tasks such as sitting and standing, the transition period begins from the subject starting to switch the task and ends when the subject completely sit/stand. A transition is missed if no correct transition decision is made within the defined transition period.

3) *Prediction Time of the Transitions:* The prediction time of a transition is defined as the elapsed time from the moment when the decisions of the classifier changes locomotion mode to the critical timing for the investigated task transitions. For the transitions between walking on level-ground and staircase (W→SA, SA→W, W→SD, and SD→W), the critical timing is defined as the beginning of the swing phase of the prosthetic side in the transitional period; for the transition ST→W, the critical timing is chosen as the beginning of the swing phase (prosthetic leg toe-off); for the transition W→ST, if the last standing leg was the prosthetic leg, the beginning of initial double limb stance phase was used as the critical timing; if the last standing leg was the sound leg, we defined the critical timing at the beginning of terminal double stance phase. For the transition S→ST and ST→S, the critical timing is the moment that the pressure under the gluteal region of the subject starts to drop to zero reading or exceed the zero reading.

RESULTS

The intent recognition system was tested on one patient with transfemoral amputation. For the studied five tasks, the overall classification accuracy in static states across 15 real-time testing trials is 98.25%. For all the 15 trials, none of the mode transitions was missed during the defined transition period. The prediction time for 8 types of transitions is shown in Table 1. This result showed that the user intent for the locomotion transitions can be accurately predicted about 76-295 ms before the critical timing for switching the control of prosthesis.

Table 1. Predication time of mode transitions before critical timing

Transition	W → SA	SA → W	W → SD	SD → W	W → ST	ST → W	ST → S	S → ST
Estimation Time (ms)	126.7 ± 28.6	136.5 ± 25.7	138.8 ± 30.5	108.3 ± 27.4	92.8 ± 35.6	127.6 ± 25.3	295.6 ± 40.8	76.2 ± 22.8

The real-time intent recognition result in one representative trial is shown in Fig. 4. During the 56 second real-time testing, totally four decision errors in static states were observed when the subject performed the stair descent task. These four errors were misclassified as level-ground

walking. All the transitions are correctly recognized before the defined critical timing within the transition period.

DISCUSSION

Similar to the experimental results observed in our previous able-bodied subject testing, the designed intent recognition system produced a 98.25% accuracy in static states and 108-138ms transition prediction time (for W→SA, SA→W, W→SD, and SD→W), although the tested amputee only has a 68% of residual limb length. This implies that the muscles in the amputee's residual limb still present different activation pattern among studied locomotion modes, which can be potentially used for neural control of artificial legs.

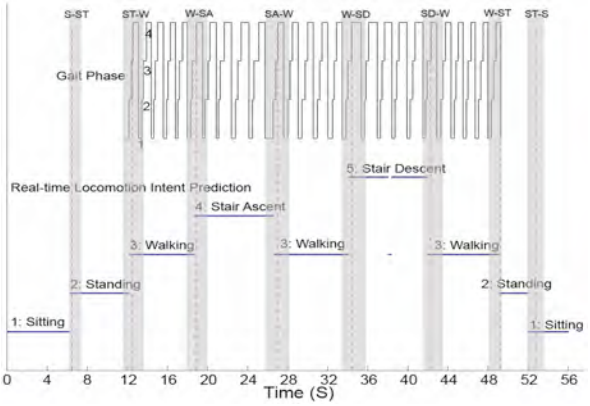

Fig.4. Real-time recognition results in one representative testing trial. The white area denotes the static states period; the gray area represents the transitional period. The red dash line indicates the critical timing for each transition.

Different from the discrete gait phases used in the previous study [10], continuous gait phases were used in this study, which makes the real-time implementation of the designed system feasible and practical. It is noteworthy that the gait phase is determined only based on the vertical ground reaction force measured from a load cell mounted on the prosthetic pylon. This design enables the system to be self-contained, which makes the integration of intent recognition system into prosthetic legs possible.

Additional efforts are needed, including (1) investigation of importance of the information carried by each sensor, (2) testing more subjects with various levels of TF amputations, and (3) study of the effects of errors of the intent recognition on the prosthetic leg control.

CONCLUSION

In this study, an intent recognition system was implemented in real-time on one patient with a transfemoral amputation. The system achieved 98.25% accuracy for indentifying the locomotion modes in static states and showed fast response time (76-295ms) for predicting the task transitions. These preliminary results demonstrated potentials of designed intent recognition system to aid the future design of neural-controlled artificial legs and therefore improve the quality of life of leg amputees.

ACKNOWLEDGEMENTS

The authors thank Zhi Dou, Andrew Burke, and Ming Liu at the University of Rhode Island, and Michael Nunnery and Becky Blaine at the Nunnery Orthotic and Prosthetic Technology, LLC, for their suggestion and assistance in this study.

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MYOELECTRIC CONTROL OF A POWERED TRANSFEMORAL PROSTHESIS DURING NON-WEIGHT-BEARING ACTIVITIES

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ABSTRACT

Lower limb prostheses have traditionally been mechanically passive devices without electronic control systems. Microprocessor-controlled passive and powered devices have recently received much interest from the clinical and research communities. The control systems for these devices rely on mechanical sensors placed on the prosthesis. Few studies have investigated control systems that rely on information extracted from myoelectric signals to help control lower limb prostheses. In this paper we show that sagittal plane motions of the knee and ankle can be accurately (>90%) recognized using only myoelectric signals (MESs) measured from residual thigh muscles. The control system for a powered transfemoral prosthesis was modified to accept myoelectric control information and subjects demonstrated real-time control of the knee and ankle for non weight bearing motions. . This research is the first step in our long-term goal of implementing myoelectric control of lower limb prostheses during both weight-bearing and non-weight-bearing activities for individuals with transfemoral amputation.

INTRODUCTION

Lower limb amputation is a major cause of disability for millions worldwide. A variety of mechanically passive prostheses have traditionally been used to restore mobility to these individuals. Microprocessor-controlled variable damping knees have recently gained popularity due to their ability to enhance knee stability and adapt to different ambulation speeds [1]. However, these prostheses still only dissipate mechanical power—they cannot generate the power required for many activities, such as standing from a chair or ascending stairs. Microprocessor-controlled powered prosthetic legs have recently become commercially available, and several prototypes are in various stages of development. High-level state-based controllers interpret signals recorded from mechanical sensors embedded in the prosthesis or from an orthotic placed on the sound limb. These signals provide control information to lower-level position, force, torque, or impedance controllers.

Myoelectric control for lower limb prostheses is a developing field of research. Recent studies demonstrate that myoelectric signals (MESs) from the residual thigh of a transfemoral amputee can be used to estimate the subject's ambulation mode activity during weight-bearing situations [2]. Using pattern recognition techniques, residual thigh muscle activity can also provide information to control a prosthetic knee [3] or a combined knee and ankle [4]. Subjects in these previous studies were not wearing prostheses during testing; the prostheses were either attached to a laboratory benchtop or the experiments were completed within a virtual environment. In this study we expand the number of subjects tested in [4] and report results for subjects fitted with a motorized transfemoral prosthesis.

METHODOLOGY

Two experiments were completed between September 2009 and May 2011 at the Rehabilitation Institute of Chicago. The Northwestern University Institutional Review Board approved the studies, and written informed consent was obtained from all study subjects.

Experiment 1: Real-Time Non-Weight-Bearing Control within a Virtual Environment

Eight subjects with transfemoral amputations (5 males, 3 female, mean (SD) age 49years, mean number of years post amputation 19 years) participated in this study. Subjects were seated and the following nine muscles were identified based on anatomical location and palpation: semitendinosus, sartorius, tensor fasciae latae, adductor magnus, gracilis, vastus medialis, rectus femoris, vastus lateralis, and long head of the biceps femoris. Nine adhesive, gelled silver–silver chloride electrode pairs were placed over the muscles of interest with an interelectrode spacing of approximately 3 cm. All data were amplified by a factor of approximately 1000, digitized using a 16-bit analog to digital converter, and transferred over a controller area network (CAN) bus using the Prosthesis Device Control Protocol [5].

Custom software—Control Algorithms for Prosthetic Systems (CAPS)—instructed the subjects to perform the following movement: knee flexion, knee extension, ankle

plantar flexion, ankle dorsiflexion, and no motion. The order that the trials were collected in was not randomized and eight repetitions of 3 s each were collected for each motion. Data from repetitions 1–4 were used to train a pattern recognition system, and data from repetitions 5–8 were used to compute classification accuracy. The pattern recognition system was based on time-domain features extracted from 250 ms overlapped analysis windows and classified by a linear discriminant analysis classifier. This system has been well-documented [6] and shown to provide good classification performance for upper limb amputees [7].

After the pattern recognition system was trained, subjects completed a motion test within a real-time virtual environment [7]. The motion test required subjects to replicate motions displayed on a computer screen while real-time position feedback was provided by a virtual avatar. Each motion test consisted of nine trials of each of the four movements (the no motion class was not tested) presented in random order. A trial was completed successfully when the subject moved the virtual limb through its complete range of motion for the tested class. Trials could be completed in a minimum of 1 s and were terminated after 15 s. Performance metrics included classification accuracy, motion completion time, and motion completion percentage [4]. Motion completion time is the elapsed time from movement onset until the virtual limb is moved through the complete range of motion. Motion completion percentage is the number of successfully completed motions divided by the total number of trials.

Experiment 2: Real-time Non-Weight Bearing Control with a Powered Knee Prosthesis.

Two of the eight participants returned to complete a second experiment to evaluate their performance when controlling a powered knee prosthesis. MES control site locations were marked on a custom fabricated socket at the end of experiment 1 and stainless steel dome electrodes were embedded into the socket wall. MES data were amplified by a factor of 1000, sampled by a 16 bit analog-to-digital converter and streamed across a CAN bus to CAPS software.

The powered knee prosthesis used in this experiment was designed and fabricated at Vanderbilt University and is similar to the prosthesis described in previous work [3, 8] except that the ankle actuation unit was removed (Figure 1).

Figure 1: Subject wearing the powered knee prosthesis

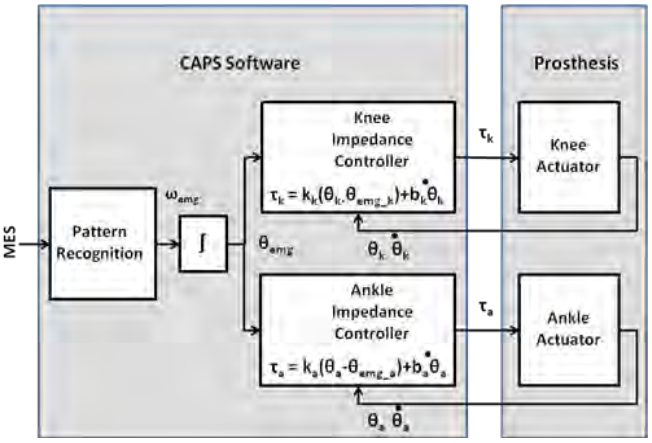

Figure 2: Architecture of the impedance controller used to generate the torque command provided to the powered knee prosthesis.

The powered knee was modified to implement the Prosthesis Device Control Protocol [5] so that it could send sensor data to and be controlled by CAPS software.

A volitional impedance controller was created within CAPS (Figure 2) and was very similar to architecture described previously by the Vanderbilt Group [3]. The pattern recognition system described in experiment 1 provided the two mutually exclusive outputs ω_{k_emg} and ω_{a_emg}, corresponding to knee and ankle velocities, respectively. These velocities were integrated to provide an estimate of the desired knee and ankle positions. A joint torque command was generated according to the following equation:

$$\tau_i = k_i\left(\theta_i - \theta_{i_{emg}}\right) + b_i\dot{\theta}, \qquad (1)$$

where i was an index corresponding to the knee or ankle, k was an empirically determined virtual stiffness, θ was the position measured from the prosthesis, θ_{emg} was an estimate of the desired joint position, b was an empirically determined

virtual damping term, and $\dot{\theta}\dot{\theta}$ was the joint velocity measured from the prosthesis.

When the prosthesis was initially powered on, the tuning parameters (k and b) were set to 0 such that a joint torque command of 0 Nm was sent to the device while training and testing data were collected. The data were collected using the same procedure as used in experiment 1. The pattern recognition system was trained to recognize knee flexion, knee extension, ankle plantar flexion, and ankle dorsiflexion, and no motion. Next, the myoelectric impedance control parameters were tuned empirically. The values of k_k and b_k were slowly adjusted until the subject could move the knee through the full range of motion at a comfortable speed with a smooth kinematic profile. Since the prosthesis did not contain an ankle actuation unit, the ankle tuning parameters, k_a and b_a, were left at 0. These parameters would also need to be adjusted in order to control an ankle actuation unit.

Table 1. Virtual prosthesis performance metrics (n = 8)*.

	Classification Accuracy (SD), %	Completion Time (SD), s	Completion Percentage (SD), %
Overall	92.1 (3.7)	2.40 (0.82)	97.6 (3.1)
Knee	92.1 (2.8)	2.03 (0.84)	100.0 (0.0)
Ankle	88.1 (9.2)	2.79 (1.23)	95.1 (6.3)
No Motion	99.9 (0.4)	n/a	n/a
		n/a	n/a

Subjects practiced controlling the knee for several minutes prior to completing motion tests with the physical prosthesis.

The motion tests were very similar to those described in experiment 1 except that the order of motions was not randomized; knee flexion and extension were tested first. Subjects were cued by the experimenter to perform the appropriate motion and move the knee joint through the full range of motion. Ankle motion tests were completed with the prosthetic knee positioned at 90 degrees of knee flexion (i.e. neutral position when sitting) and at 45 degrees of knee flexion. Testing in the two different positions allowed us to determine if the pattern recognition system could still recognize ankle motions when the knee was repositioned. Feedback was provided to the subject by both the virtual environment and the physical prosthesis: the output of the pattern recognition classifier was displayed on a computer monitor and if the pattern recognition system erroneously decoded a knee command, then the prosthesis would move. The performance metrics of the motion tests were motion completion percentage and motion completion time.

RESULTS

Experiment 1: Real-Time Non-Weight-Bearing Control within a Virtual Environment

Subjects achieved high classification accuracies and completion percentages for both knee and ankle motions (Table I, Figure 3). The classification accuracy from one of the subjects was excluded as an outlier; we determined that this subject only held the contraction briefly while training data were collected resulting in many 'no motion' class errors. Nonetheless, this subject could still control the prosthesis during the real-time tests.

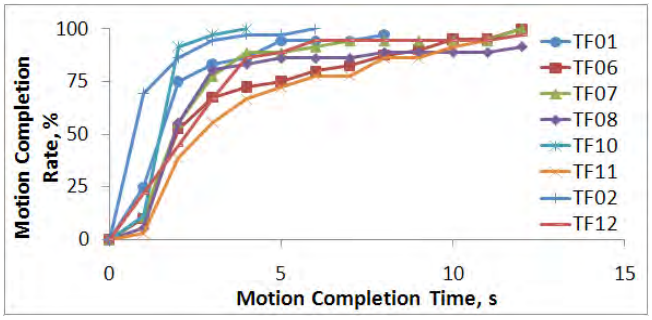

Figure 3: The cumulative motion completion percentage foreight subjects.

Table II: Comparison of Physical and Virtual Prosthesis Performance Metrics (n - 2)

	Virtual Prosthesis	Physical Prosthesis
Classification Accuracy (SD), %		
Overall	• 92.5 (0.7)	• 94.5 (3.5)
Knee	• 93.3 (1.8)	• 93.8 (2.5)
Ankle	• 89.3 (0.4)	• 93.0 (5)
No Motion	• 100 (0)	• 100 (0)
Completion Time (SD), s		
Overall	• 1.27 (0.2)	• 1.30 (0.1)
Knee	• 1.05 (0.1)	• 1.12 (0.1)
Ankle (all)	• 1.48 (0.4)	• 1.35 (0.1)
Ankle (knee at 90 deg)	• n/a	• 1.53 (0.1)
Ankle (knee at 45 deg)	• n/a	• 1.23 (0.3)
Completion Percentage (SD), %		
Overall	• 97.2 (3.9)	• 96.3 (5.2)
Knee	• 100.0 (0)	• 100.0 (0)
Ankle (all)	• 94.4 (7.9)	• 95.8 (5.9)
Ankle (knee at 90 deg)	• n/a	• 100.0 (0)
Ankle (knee at 45 deg)	• n/a	• 88.9 (15.8)

Experiment 2: Real-time Non-Weight Bearing Control with a Powered Knee Prosthesis

The tuned impedance parameters were $k = 0.8$, $b = 0.05$ for subject 1 and $k = 0.6$, $b = 0.08$ for subject 2.

Subjects performed slightly better with the physical prosthesis in comparison to using only the virtual environment (Table II, Figure 4). Importantly, the pattern recognition system could still reliably decode ankle motions when the knee joint was repositioned at a 45 degree ankle (Table II).

DISCUSSION

Accurate classification of knee motions was expected because the MESs were recorded from physiologically appropriate residual limb muscles that had previously been used to control the knee. Accurate classification of ankle motions was unexpected; the muscles that control the ankle are located below the knee and were lost as a result of the amputation. Nonetheless, subjects were generating distinct and repeatable co-activity patterns that were properly interpreted by the pattern recognition system.

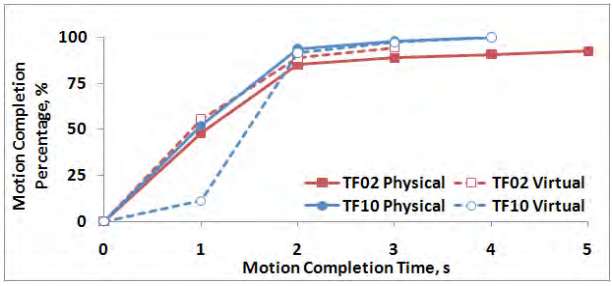

Figure 4: Cumulative motion completion percentage comparing performance between the physical prosthesis and the virtual prosthesis for two subjects, TF02 an dTF10.

This is analogous to recognizing subtle differences in hand grasp patterns using only the extrinsic forearm muscles of transradial amputees [9].

To the authors' knowledge, this is the first demonstration of myoelectric control of a powered transfemoral prosthesis. Although the results are preliminary, they are promising. Both subjects were able to reliably to control the knee in real time. Furthermore, the pattern recognition system properly interpreted ankle commands when the prosthesis was repositioned to a 45 degree angle, suspending freely in space from the socket. This suggests that MES changes resulting from dynamic loading on the socket do not degrade pattern recognition performance. Further testing with additional amputees is required to see if this result can be generalized across subjects. It also should be noted that only changes in the knee angle were tested and not changes in the position of the residual limb.

Proportional control estimates of knee velocity were not incorporated into the control system, and the parameters of the myoelectric impedance controller were adjusted empirically by the experimenter. Proportional control signals may be added by taking a simple average of MES amplitudes [10] or by using a weighted average of MES amplitudes determined by principle component analysis [3]. Smoother kinematic profiles may be obtained by optimizing the selection of the impedance parameters—the objective of ongoing research.

FUTURE WORK

Non-weight-bearing control is only one portion of the overall control system for a powered lower limb prosthesis. Non-weight-bearing control may be considered an activity mode in a state machine constructed to control the prosthesis during both weight and non-weight-bearing situations (Figure 5).

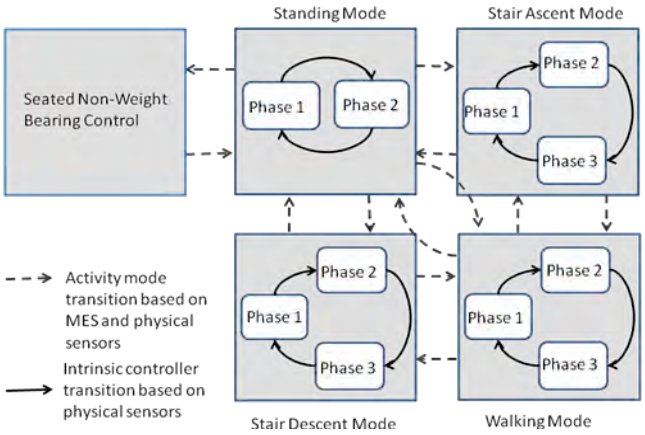

Figure 5: Conceptual block diagram of the overall control system for a powered prosthesis.

Existing powered lower limb prostheses use mid-level 'intrinsic controllers,' depicted conceptually inside the square boxes in Figure 5, to generate appropriate joint torques that are sent to the prosthesis [11]. Current intrinsic controllers rely on mechanical sensor data to transition between phases. Mechanical sensor data is also currently used to transition between activity modes. MES data has been shown to provide information that helps discriminate between activity modes [2]. Future work will quantify the benefits of adding MESs to improve activity mode recognition rates and reduce latencies between activity mode transitions.

ACKNOWLEDGEMENTS

This work was supported by the US Army Telemedicine and Advanced Technology Research Center (TATRC) project W81XWH-09-2-0020. The authors acknowledge Ann Barlow and Aimee Schultz for editing the manuscript and Todd Kuiken for project guidance.

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MOTOR CONTROL PROCESSES WHEN LEARNING TO USE A PROSTHETIC DEVICE

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INTRODUCTION

Prosthetic devices are designed to increase the action possibilities of an amputee. Appropriate actions with upper-extremity prostheses are only possible when these devices can be controlled dexterously. Importantly, the control signals of the neuromotor system necessary to perform a goal-directed action with a prosthesis differ from those control signals used to perform an action with an intact limb. To discuss what it means for the neuromotor system to learn to control an upper limb prosthetic device, the current presentation will start from Bernstein's (Russian original from 1940, published in English in 1996) insightful treatise on the hierarchical levels for the control of movement. From this overview we aim to make recommendations regarding the issues that research on learning to control a prosthetic device for the upper extremity should focus on.

LEVELS OF CONSTRUCTION OF MOVEMENT

Bernstein was a neurophysiologist with a primary interest in motor control. Based on the evolution of the brain he distinguished four levels of control of human movement that each controlled a different class of movements. These levels were hierarchically organized with each level emerging on top of the existing levels. Each new level emerged based on evolutionary pressure requiring a new class of movement. More specifically, based on new challenges in the environment, new actions had to evolve to meet these challenges. These newly evolved actions were accompanied with new sensory correction mechanisms. Based on the interplay between the newly emerged actions and the accompanying sensory corrections, new neural structures evolved. These new neural structures represented a new level of construction of movement and accounted for a new class of movements. Importantly, in evolution, and, thus, presumably in motor learning, motor functioning and sensory functioning developed mutually.

The four levels Bernstein (1996) distinguished in motor control were: the level of tone, the level of synergies, the level of space and the level of action. The lowest level of motor control, and also the oldest, is that of tone. This level controls the background muscular force that provides postural stability underlying all acts. The next level is the one that emerged when extremities evolved; it controls the linking together of muscle-articular groups so that the numerous muscles become controllable to perform stable and reproducible movements. Note that sensory corrections were primarily based on proprioception at the level of tone and that of synergies. The sensory corrections at the other two levels are primarily visually based. The level of space regulates that movements reach their goals in the workspace surrounding the body; distances and orientations of objects are perceived for reaching movements to be goal-directed. The highest level of control is that of action, in which sequences of movements are controlled. This level of control takes care of adaptive solutions to new situations. The levels of tone and that of synergies are the supportive levels whereas the level of space and action take the leading role.

LEVELS AND PROSTHETIC USE

Level of tone

This level of motor control is easily overlooked since it functions in the background in daily activities. However, for prosthetic control it is important to take this level into account. Note that the distribution of the mass of the body changes after an amputation. This affects the harmonicity of the walking pattern of an unilateral amputee. Wearing a prosthesis partly reduces such disturbances of walking patterns (Bertels et al., 2010). However, the mass and mass distribution of an upper limb prosthesis is different from that of a sound arm. This implies that the moments and forces around the proximal joints, such as the shoulder joint differ with a prosthesis compared to that of a sound arm. This will disturb postural control. Hence, optimizing a prosthesis should take such issues into account. Moreover, training programs may have to focus on how to incorporate the use of preparatory muscle activity that counteracts the forces that the prosthesis produces.

Level of synergies

Together with the level of space, the level of synergies is most important to consider when learning to use a prothesis. Active upper extremity prostheses can be broadly

distinguished into those that are controlled with myo-signals and those that are body-powered, using a harness. The use of both types of prostheses implies the learning of different synergies.

The definition of synergies employed in the current paper is the one proposed by Bizzi and d'Avella (Bizzi et al., 2008; d'Avella et al., 2006). Muscle synergies are defined as a distinct pattern of activity over time of a group of muscles. To produce a movement a set of synergies is combined. To execute different movements the onset time and amplitude scaling of each individual synergy are adapted.

For myo-electrically controlled prostheses it is important to understand that the myo-signals picked up by the prostheses' electrodes are essentially the muscle activation patterns that result from the activation of synergies. Hence, learning to control a myo-electric prosthesis implies learning to activate the appropriate set of synergies and scale the onset timing and amplitude in an appropriate way. Note that in a myo-electric prosthesis usually the flexors and extensors of the wrist are activated to control closing and opening of the prosthetic hand. These are different muscles than are used in the sound hand, hence, learning to use a prosthesis implies learning new scaling parameters of the synergies comprising these muscles.

The same is true for learning to control a body-powered prosthesis. However, now different synergies have to be learned. Usually the contra-lateral shoulder is used to control the hand opening of the prosthetic hand, hence, it is obvious that the muscles involved for the prosthetic hand control are different from those that are used in a sound hand. Thus, in this case the synergies around the contra-lateral shoulder need to be scaled to produce the appropriate muscle force that controls prosthetic hand opening. Importantly, the sensory corrections of the body-powered prosthesis differs from that of the myo-electrically controlled prosthesis. The body-powered prostheses can be controlled with proprioception signals because of the forces required to control them. This might result in a relatively easier learning of the new synergies controlling the prosthesis because these synergies are regulated mainly by proprioception.

Surprisingly, at the moment no studies are available in the literature addressing the change in activation of synergies when learning a new motor task. Moreover, the idea that the myo-signals picked up by myo-electric prosthetic devices result from synergies is in line with recent developments in the design of pick-up mechanisms in that multiple electrodes are used and that pattern recognition algorithms are used to detect a larger ranges of choices to control more complex prosthetic hands.

Level of space

Goal-directed reaching and grasping are controlled at the level of space. Reaching and grasping is done differently with a prosthesis than with a sound hand: (i) the grasp takes longer and has a relatively long decelerative phase, (ii) the grasp starts after the reach has been initiated, and (iii) the grasp profile shows a plateau phase (Bouwsema, et al., 2010; Wing and Fraser, 1983). One of the reasons for these deviations in the grasping profile might be the lack of proprioceptive feedback about the prosthetic hand. This makes that prosthesis users have to rely solely on visual feedback for aspects of motor functioning while in the sound hand proprioception is a primary source of feedback. Training of prosthesis' use should focus on making the grasp with a prosthesis more fluent. Moreover, technical developments should concentrate on providing more and appropriate sensory feedback about the prosthesis in use.

Level of action

The highest level of control of movement regards the control of sequences of actions. Two issues will be discussed. First, in sound behaviour the gaze usually precedes the manipulative actions of the hand, that is, when picking up an object the gaze usually arrives at the object before the hand and the gaze never checks the hand. The presentation of Bouwsema et al. (2011) will address how this is done with prostheses. Second, to manipulate an object, that object needs to be oriented relative to the hand in an appropriate way. This means that actions need to be planned in advance, so that the object is grasped in such a way that is appropriate for the upcoming action. This implies that during object manipulation switches between grasps have to be made. This requires prosthesis that allow for a swift alteration between hand posture, which require low attentional costs.

This latter aspect is especially important for the recently developed multi-articulated prosthetic hands. Because more grip patterns are available with these hands, the grips can be adjusted to details of the function of the hand at each particular moment in a task performance.

CONCLUSIONS

We presented a view on motor control that allows to frame the problems of research into the use of upper extremity prosthetic devices in one framework. This framework allows for a hierarchical approach of the problems of prosthetic use. From this view the change between different grip patterns should be made easy, the feedback about the prosthesis should improve, the reaching and grasping should be more fluent, training should take into account that new synergies have to be learned, and the postural disturbances following prosthetic use should be considered in training and in developing prosthetic devices.

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THE EFFECT OF VISUAL BIOFEEDBACK FORMS OF MYOELECTRIC SIGNAL ON MYOELECTRIC SIGNAL SEPARATION TRAINING

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INTRODUCTION

Myoelectric control of powered upper limb prosthesis enables the user to control the timing and amount of joint movement of the prosthetic component[1]. However, the lack of tactile sensory feedback in the control loop causes difficulties in fine control of the prosthetic component[2]. With the natural learning ability, the amputee can form a predictive control, or acquire a control method on incidental feedback during the practical use of the prosthesis. Therefore, an amputee willing to use any kind of myoelectric controlled prosthesis is strongly recommended to spend time to practice with the system to achieve the intended task. It is, therefore, important to enhance the learning processes of embodying the donned prosthesis. Nevertheless, the user of a myoelectric control system is fundamentally required to voluntary alter the myoelectric signal. It is import to conduct research on the learning process of myoelectric control and the evaluation and feeding back method for the trainer and trainee. In this paper, a quantitative evaluation method for scaling the degree of separation of the myoelectric signals is presented and an experimental system for myoelectric signal isolation training is developed to test the effect of four types of visualization methods.

EXPERIMENTAL SYSTEM

A Personal computer based myoelectric tester is developed for visualizing the myoelectric signal or the detected movement from the signal, as in a commercially available system and previous researches[1,3]. The target of this research was focused on the phase of myoelectric assessment and early stage of myoelectric signal control training. In this phase, the target of evaluation and training is improving the independent control and relaxation of flexion and extension muscle to reduce the co-contraction. Therefore, the main function of the software is to visualize the wavering myoelectric signal in realtime for self modulation.

Four types of graphical forms were prepared for the experiment. The forms were selected from the commercially available system, waveform, bar graph, and animation of the computer graphic, CG, hand. In addition, planar distribution graph of the 2-site myoelectric signals, which is used in describing the relation of activity level of the 'operating points' in the proportional control[1], is also prepared. The forms' screen shots are shown in Figure 1.

The experimental system consists of 2 sets of myoelectric sensors (Otto Bock, MyoBock 13E200=60) and a personal computer with an AD converter board (Interface, PCI-3168). The sampling frequency is set to 1 kHz. The raw data sampled from the sensor signal is concurrently recorded. In the CG form, two-site two-function On/Off control strategy under first-come-first-served condition is used to control the hand opening/closing.

The tester was designed with two modes, practice and evaluation. In the practice mode, one out of the 4 types of screen is shown on the 22-inch LCD. Targeted level band, which the subject tries to hold the signal within, is shown in diluted color. The target band are shown, 'On,' for 2-seconds and 'Off' 3-seconds in series, while the sites and the levels, high and low, are switched consecutively. The On/Off is repeated 20 times for one set of trial, and 5 sets of trial is carried out with intervals in between. For the CG form, the target finger positions are shown in blue. Once the operating finger is located within the allowable displacement and maintained for 3-seconds, the next target position is shown. This is repeated 20 times as one set of trial. As for the evaluation mode, the CG form is shown and 5 sets of the above mentioned routine is carried out.

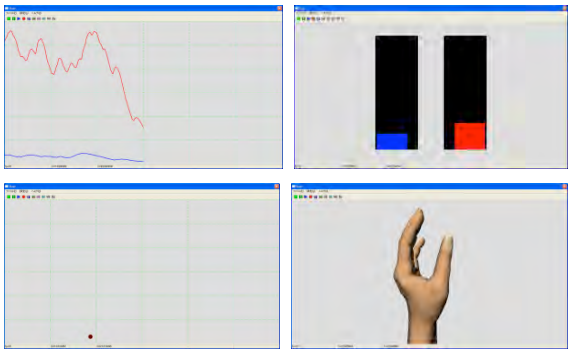

Figure 1: Forms for the practice and evaluation modes. From top left, waveform, bar, planner distribution, and CG.

EXPERIMENT

The experiment was approved by the university IRB and all subjects participated in the experiment after providing a written informed consent. 20 non-amputated adult males (Mean: 23.2 year-of-age) participated. The myoelectric sensors were placed on the surface of their non-dominant forearm. The initial positions of the sensors were selected after palpation and confirming the muscle contraction. The sensors were attached to the skin with adhesive tape and an elastic strap was wrapped over the sensor. Subjects sat on a chair in front of the monitor and the tested arm was held at elbow flexion 90 degrees during the practice. All subjects participated only once in a 5-day program: consisting of 5-sets of pre-practice evaluation, 5-sets of practice and 5-sets of post-practice evaluation, each day. Five subjects, randomly selected, took part in the forms, respectively. The subject's wrist joint were braced with a plastic cast and fixed to a posture of the thumb positioned upward.

To quantitatively evaluate the quality of the isolation during the voluntary muscle contraction, the next equation is applied. The degree of isolation D_t is calculated from the flexor myoelectric singal V_{Ft}, and extensor myoelectric signal V_{Et}, sampled at time t. Since the signal isolation is important at activation, the period which one of the signals exceeded the threshold of 0.2 Maximum Voluntary Contraction is selected from the recorded myoelectric signals and computed. This threshold is also used in the CG movement discrimination algorithm.

$$D_t = \frac{|V_{Ft} - V_{Et}|}{V_{Ft} + V_{Et}} \quad (V_{Ft} \neq V_{Et} \neq 0) \qquad (1)$$

RESULT

The average degree of isolation computed for each subject for the pre-practice evaluation of the first day and the post-practice evaluation of the last day is shown in Figure 2. Nineteen subjects, out of 20, had higher isolation at the end of their training period. Multiple-conparison, Bonferroni t-test, was conducted to analyze the influence of the visual forms used in the practice. The subject's last day's post-practice evaluation result was tested. As described in Table 1, the result showed no significant differences between the groups.

DISCUSSION

With the variety of subject's evaluation results show that the degree of isolation has good sensibility as a scale to detect the changes of individual's performance.

The statistical analysis results show that the screen setting and the properties of visual information fed back to the subjects were not the major factors for varying the degree of isolation. No pair of comparison showed notable difference and this was confirmed to be equivalent for the evaluation results of the pre-practice result of the first day. From these result, it can be assumed that the practice can be planned on any form. Finally, caution is necessary since the results are extracted from a limited subject population. All subjects in this experiment major in engineering and are keen of graphical representation of collected data. The planner distribution form may be difficult to appreciate in some amputee, and further testing is essential.

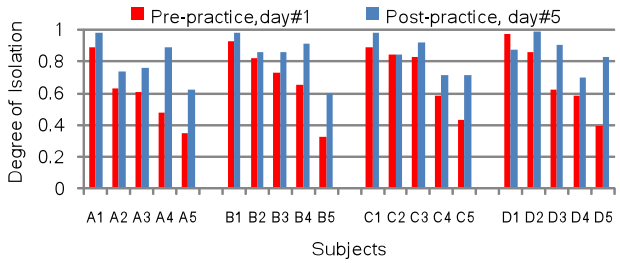

Figure 2: Degree of isolation of the 20 subjects. The subjects in Group A worked on the waveform, group B bar graph, group C planner distribution, group D CG.

Table 1: Multiple test result of comparing the effect of visual forms on the degree of isolation.

Factors	ANOVA				Multiple Comparison
	df	F	η^2	p	
Visual Inf.	3	0.197	0.036	0.90	NS
Error	16	(0.016)			

CONCLUSION

A Personal computer based myoelectric tester is developed. Four types of graphical form were prepared for monitoring the signal during the practice and virtual reality hand is used for the pre- and post-practice assessment. A numerical function to evaluate the isolation is proposed. Experiments on 20 subjects were conducted. The statistical analysis found that there is no significant difference between the graphical forms of the signal in practicing the isolation.

ACKNOWLEDGEMENTS

This research experiments were supported by Okayama Prefectural University. The authors thank Dr. Takaaki Chin, Hyogo Rehabilitation Center for the clinical advises.

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TRAINING INDIVIDUALS TO USE PATTERN RECOGNITION TO CONTROL AN UPPER LIMB PROSTHESIS

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INTRODUCTION

Pattern recognition has been used in the laboratory for control of advanced prosthetic limbs. However, recent work has shown that it has the potential to improve control of existing clinical prostheses [1]. Targeted muscle reinnervation (TMR) makes it possible to access neural information from residual peripheral nerves that previously innervated the missing limb [2]. Pattern recognition allows this information to be extracted and used by individuals with high-level amputations for effective prosthesis control. TMR is not necessary for pattern recognition control at the transradial amputation level [3]. Here, we outline a sequence for training individuals to use pattern recognition control of elbow movement, wrist rotation, wrist flexion/ extension, and hand grasps. We highlight the differences between training for direct control and training for pattern recognition control. We also recommend a training protocol to facilitate mastery of pattern recognition control before and after being fitted with a prosthesis.

TRAINING PROGRESSION

Teaching the *concept* of pattern recognition control

Understanding pattern recognition control is the first challenge for individuals with an amputation We begin with verbal explanations of pattern recognition, including the fact that each electrode location no longer corresponds to a specific movement (as in direct control), and that consistent patterns of muscle activations are required for each movement. We encourage the individual to actively participate in training. The process of selecting shared vocabulary such as "channel," "signal," "degree of freedom," "supination," or "key pinch," engages the individual as a partner in the process and invites their active participation. Agreed-upon terminology also ensures clear communication between the individual and the clinician.

Once electrode sites are located (typically 6 for the transradial level and up to 12 for the shoulder disarticulation level), we use a myoelectric signal viewer that shows patterns of myoelectric activity corresponding to movement attempts. This illustrates to the individual how they are able to produce identifiable patterns of muscle activity for a given movement. We are also able to use virtual reality software to provide feedback to the individual as they attempt control.

We explain the importance of performing the intended movements with a moderate level of effort to avoid fatigue, and the necessity of duplicating the level of effort for each movement, as a significant change in effort may confuse the classifier. Frequent retraining of the classifier is performed in initial training sessions because of physiological changes, such as altered skin conduction, or alterations in movement attempts that occur as the individual adapts to the training process. It is explained to the individual that retraining is expected and will be part of the routine, although it may become less frequent as they gain experience in using pattern recognition control.

Phantom limb considerations

It is appropriate early in the training process to discuss the role of the phantom limb in pattern recognition. It is necessary to determine if the phantom limb will be useful during training. Users should be instructed to try to move their phantom limb in the desired direction, even if it feels immobile. Some individuals experience pain or cramping when attempting to move the phantom limb; this discomfort may interfere with successful control. In this case, we instruct the individual to use a moderate level of effort, to focus mirroring the desired movement with the intact limb, and to allow time for relaxation of the phantom limb. Photographs of exercises to be performed with the phantom limb are useful and can be included in a home exercise program in preparation for the next training session (Figure 1).

Pattern recognition training for individuals with transradial amputation

Initial training sessions utilize a virtual arm with which the individual first experiences pattern recognition control. We begin pattern recognition classifier training with the degrees of freedom easiest for the individual to control—hand open and close for the transradial level. Individuals with transradial amputation must learn new motor commands: in pattern recognition control, the individual is being asked to perform a movement that he or she is not accustomed

to controlling *intuitively*. With direct control, wrist flexion and extension are used to open and close the hand, whereas physiologically appropriate muscles are used in pattern recognition control. Frequent rest breaks may be needed as fatigue is common when learning new muscle activation patterns or when a myoelectric prosthesis has not previously been used.

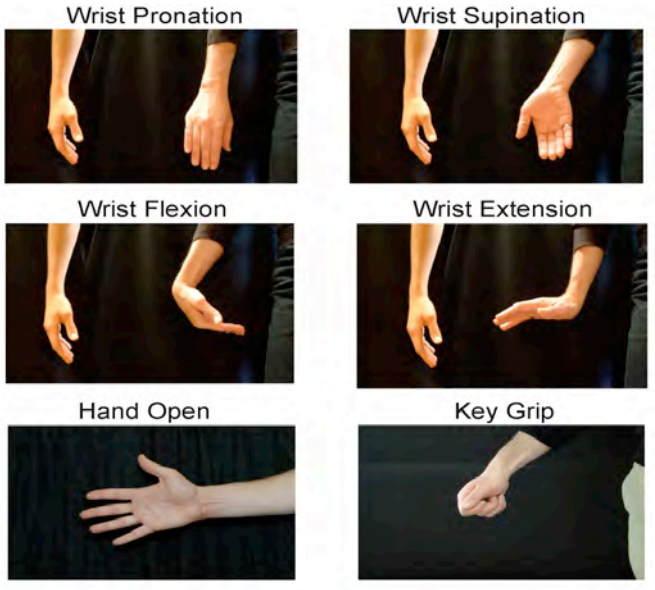

Figure 1: Examples of phantom limb exercises for home program

Next we try adding degrees of freedom outside of the individual's experience. For individuals with transradial amputations, these are wrist flexion/extension and multiple grasps. This provides an opportunity to demonstrate to the individual the potential for pattern recognition to enhance prosthesis functionality and to highlight the amount of motor-control information available in the residual limb. The concept of 'retraining' should be discussed again with the individual, as retraining the classifier is indicated whenever control seems degraded or a new degree of freedom is added. Activities that demand an extreme arm position, such as overhead reach, may also require retraining.

Pattern recognition training for individuals with transhumeral or shoulder disarticulation amputations and TMR

Individuals with higher-level amputations and TMR begin classifier training in the virtual reality environment with elbow flexion and extension. We progress quickly to an added degree of freedom, usually hand open and close, which is available with direct control in conventional myoelectric prostheses. Because there is more control information available in the residual limb after TMR, pattern recognition affords intuitive control of two degrees of freedom at the wrist and multiple grasps. Activating three or more degrees of freedom demonstrates to the individual the potential of pattern recognition control. At the end of the first session, it is appropriate to send the individual home with a program to exercise new motion attempts with their residual and phantom limbs.

Subsequent visits for pattern recognition training will begin with a review of the exercise program, or any symptoms related to unaccustomed use of muscles. It is beneficial to repeat the discussion of pattern recognition concepts while setting up for the training session and to begin working with the degrees of freedom that were successful at the last visit. It may be appropriate to train and test two or three degrees of freedom using virtual reality programs.

If there are any unintended movements, some time might be devoted to distinguishing actions based on verbal information from the individual. The subject may need to describe or demonstrate the movement with the intact limb. Make clear that the pattern recognition control model does not allow for 'parallel' classification: only one decision (or movement) can be performed at a time. Also, during training, 'no movement' classes are important for distinguishing between movement classes. As degrees of freedom are mastered we add more degrees of freedom, up to the capability of the prosthesis intended for use.

Evaluation of control in the virtual reality environment

When control in the virtual environment has been mastered, control can be assessed using the Motion Test and the Target Achievement Control (TAC) Test [4]. In the Motion Test, the individual is randomly prompted to perform a single movement. The movement has to be completed within a given time frame, and inadvertent movements are ignored unless they directly oppose the requested movement (such as wrist flexion performed during a wrist extension trial). In the TAC Test, the trainee must sequentially position one or more degrees of freedom to achieve a target posture, and misclassifications must be corrected within an established time frame for the trial to be considered successful. This increases the difficulty and lets the clinician focus on the most challenging aspects of control. The time frame and tolerances for the TAC Test can be adjusted as performance improves.

Pattern recognition control of a prosthesis

As the virtual arm avatar does not change with respect to position in space, it is necessary to move away from training with virtual reality to training with a prosthesis. Controlling a prosthesis remotely (Figure 2) is a tool used during the early sessions before a socket is fabricated.

Figure 2: Remote control of prosthesis using pattern recognition

The socket is usually completed by the third or fourth session, and pattern recognition training while wearing a prosthesis can begin. Training is done with the arm supported or unsupported in approximately 45 degrees shoulder flexion. It may be necessary to collect training data both while the individual is standing and when he or she is sitting. Proximal postural effects, weight of the device, prosthesis position relative to gravity, and length of residual limb all affect direct myoelectric control and will likewise affect pattern recognition control. To train for functional prosthesis control and use, we introduce common objects for grasping and change their orientations to provide a variety of pre-positioning demands. Reminders to perform the movements in the same way as during training are helpful. The addition of the prosthesis and the introduction of increased functional demands add to the challenge. Again, it may be necessary to remind the individual that the classifier may need to be retrained.

The individual probably does not have experience using a prosthesis with powered wrist flexion and extension or multiple grasps. It is useful to guide the individual during prepositioning while they are accessing the new functions of the prosthesis instead of using customary postural accommodations (see Figure 3). Prepositioning demands increase as additional degrees of freedom are included in the classifier, until all degrees of freedom possible with the prosthesis are utilized.

The clinician should suggest alternative prepositioning techniques that use wrist flexion/extension or alternate grasp patterns to demonstrate the added potential of the prosthesis. Simple functional tasks like the 'clothespin relocation task' and The Southampton Hand Assessment Procedure [5], can be useful for developing prepositioning skills and measuring progress.

Allowing individuals to watch video of themselves using the prosthesis is instructive in demonstrating progress. Watching video of others who have mastered pattern recognition control of a similar device can also demonstrate the potential of pattern recognition control.

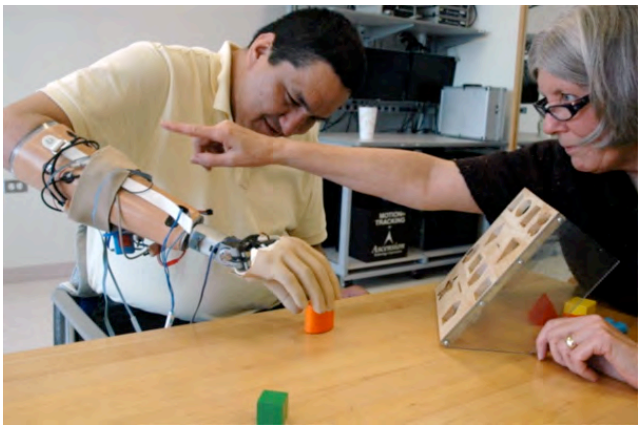

Figure 3: Postural accomodations

The next stage in training for pattern recognition control is to integrate function of both extremities. Bimanual function is essential and increases the complexity by another degree. Initially the intact limb is somewhat passive, performing a holding function. Gradually, simultaneous action using both limbs is encouraged in activities such as folding towels, using a tape measure, opening cupboards, and picking up and carrying a tray or basket. Actions can then progress to alternating limb activities such as hanging clothes, opening packets, using scissors, and cutting fruit. The individual is encouraged to decrease the amount of visual attention paid to the prosthetic terminal device.

Once the individual has reported satisfaction with the performance of the prosthesis and can demonstrate basic skills, cognitive demands in functional tasks may be increased. Tasks are given in which more organization and planning is required for successful task completion and divided visual attention is needed to perform the task in a timely manner: prepare a meal, pack a suitcase, assemble bookshelves, or sew on a button. Verbal cues to retrain may be needed if unusual positions affect control. It is also useful to take this opportunity to do an Assessment of Capacity for Myoelectric Control [6] to get a baseline score of control.

FUTURE WORK

Our experience with pattern recognition control in the home and community to date is limited. Due to upcoming software and electronic improvements, further training development will be needed. We anticipate advancing the application of pattern recognition control to the home

environment in the near future. We look forward to further refining our approach to training individuals with amputations at all levels in the use of pattern recognition control with conventional and advanced prosthetic devices.

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FUNCTIONING OF CHILDREN WITH UNILATERAL CONGENITAL BELOW ELBOW DEFICIENCY: AN ONLINE FOCUS GROUP STUDY

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INTRODUCTION

Children and adolescents with unilateral congenital below elbow deficiency (UCBED) seem to function quite well in daily life[1-3]. However, current literature does not give insight into actual functioning of these children. Furthermore, it is unknown how the children and adolescents themselves think about their functioning. Functioning encompasses activity and participation according to the Child and Youth Version of the International Classification of Functioning (ICF-CY)[4]. Activity limitations are difficulties an individual may have in executing activities. Participation restrictions are problems an individual may experience in the involvement in life situations. According to the ICF-CY, both environmental and personal factors can affect activity and participation, and thus someone's functioning.

Aim of the study:

The first aim was to evaluate whether children and adolescents with UCBED experience activity limitations and participation restrictions and if they do, how they deal with those limitations and restrictions. Secondary aims were to examine differences in activities and participation for different age groups and to compare the perspectives of children, their parents and health professionals.

Patients

Children with UCBED aged 8-12, 13-16, and 17-20 years, parents and professionals participated in the study. Participants were recruited from several rehabilitation centers in the Netherlands. Furthermore, participants were invited to take part in the study through websites of patient organizations.

METHODS

A qualitative study design was applied by using Online Focus Group interviews[5]. The interviews were held in the asynchronous form, meaning that participants could decide themselves when to log in and take part in the online discussions within a time frame of seven days. The online focus group interviews were held on a secured website containing five separate forums, one forum for each group of participants. During the first five days of the week, at the beginning of each day, a question was posted on the forums. Discussion topics were activities, participation, prosthesis use, emotional functioning and rehabilitation care. During the last two days, the participants had the opportunity to bring in their own discussion topics. The framework approach was used for data analysis.

RESULTS

878 postings were received from 17 children of 8-12 years of age, 13 teenagers of 13-16 years of age, 12 adolescents of 17-20 years of age, 17 parents and 19 professionals. Having a short arm did not prohibit execution of any activity, but not all children were able to perform all activities. The children en parents mentioned numerous creative strategies to deal with a short arm. Although people in the direct (internal) environment of the child, such as parents and friends, can be supportive, it was remarkable how often people in the indirect (external) environment of the child were mentioned as a reason for a limited functioning of a child with UCBED. People in the external environment judge a child's capacity without having sufficient knowledge about their abilities. Environmental factors were especially decisive in transitional phases, like going to a new school or applying for a new job.

Personal factors also influenced the children's and adolescent's functioning. Not all children had the same cognitive or motor abilities, react in the same way emotionally or behave in the same way in social situations. Parents were positive about the functioning of their children. Overall, parents did not think their child experienced many limitations. Health professionals described fewer strategies to deal with limitations and emphasized benefits of prostheses more than other participants.

CONCLUSIONS

Children with UCBED don't feel disabled, but environment can make them feel that way, especially in transitional phases.They have numerous strategies to deal with their deficiency. Prostheses are a minor solution to overcome limitations.

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UPPER LIMB PROSTHETIC SERVICES POST HAITI EARTHQUAKE

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HISTORY OF PROSTHETICS IN HAITI

The International Society of Prosthetics and Orthotics (ISPO) and the World Health Organization (WHO) have estimated that people needing prostheses or orthotics and related services represent 0.5% of the population in developing countries [1]. In pre-earthquake Haiti, there was a paucity of data available on persons with amputations, although it was recognized that the services available were not sufficient to meet the needs of the population [2,3]. In one survey, the most common cause of amputation was infection, followed by motor vehicle accidents, and only 25% of persons had received prosthetic rehabilitation [3].

At the time of the earthquake, Healing Hands for Haiti (HHH) operated the only full time prosthetics and orthotics laboratory, with on-site rehabilitation therapy and medical services. Six technicians had been trained in apprenticeship format through visiting expatriate CPOs, and by December 2009 participation in a credentialed training program through Don Bosco University in El Salvador [1] was in the process of being finalized through a collaborative effort of Healing Hands, Physicians for Peace and the International Committee of the Red Cross (ICRC).

Polypropylene technology with modular componentry and custom sockets was the primary type of prostheses fabricated in the Healing Hands clinic. The decision to utilize this method was based on over 10 years of experience in the country with prosthetics and orthotics services, which included long term follow up of patients, training of community based rehabilitation workers and technicians, and an evaluation of both durability, acceptability and affordability of the devices. A system that could be fitted, modified, repaired and replaced in Haiti by local technicians, at an affordable cost, was felt to be superior to alternatives such as fabrication out of country, or short-term 'full service' clinics and workshops conducted by expatriate visitors. Consultation with the ICRC was essential in providing guidance and ultimate support throughout this process.

Upper limb amputations had typically been managed with direct assistance of visiting CPO mentors, often with components being transported from USA or Canada, and fabrication and fitting occurring over a succession of visits by international CPOs. Our experience has been that cosmesis is important, and hook terminal devices were not well received. Despite progress with lower limb services, the capacity to manage upper limb needs remained extremely limited at the time the earthquake struck.

THE EARTHQUAKE AND AMPUTATIONS

On January 12, 2010 a 7.0 magnitude earthquake struck near the capital of Haiti, a Caribbean nation typically referred to as "the poorest country in the Western Hemisphere". The devastation was profound, and with over 300,000 injured the need for urgent and emergent rehabilitative services was paramount [2,4]. Early estimates of over 2000 persons newly amputated as a result of injuries and secondary complications [2], coupled with significant media attention, led to an unprecedented international response, with over 20 organizations pledging support for prosthetic services. Coordination of these actors represented significant challenges, and many groups did not seek to collaborate with the World Health designated leads for Rehabilitation, or with those providers already operating in the country pre-earthquake. This led to many types of fabrication techniques, prosthetic components and service delivery models, including some that functioned exclusive of any national provider. HHH partnered with Handicap International as the HHH clinic was destroyed, and began operation of the joint Physical Rehabilitation Centre (PRC) in February 2010. Pre-earthquake local staff were joined by expatriate volunteers and staff to continue provision of limbs with polypropylene technology, along with rehabilitation therapy, delivered by local staff complemented by expatriate mentors. By June 2010, over 200 patients received limbs at the clinic, and nation-wide an estimated 600 patients had been fit between the various providers.

Ultimately, the estimated numbers of persons with earthquake-related amputations in Haiti was revised to 1200-1500, and by approximately 6 months post-earthquake, over half of those had received a lower limb prostheses, a response not seen in any recent natural disaster of this magnitude.

Upper limb loss was not prioritized by any of the organizations, although the HHH/HI clinic (PRC) initiated evaluations for need and type of upper limb prostheses in the 12 weeks following the earthquake. Once the emergency phase of services provision was completed, introduction of upper limb fitting and technician training began. We present here the early results of the upper limb amputation program provided by Healing Hands for Haiti and Handicap International, in the aftermath of the 2010 earthquake.

Table 1: Breakdown of levels of amputation observed in 107 consecutive patients with amputations examined in 17 field hospitals between days 3-17 post-earthquake. Amputations made up 35% of in-hospital injured in need of immediate physical rehabilitation. N=307 consecutive patients [2].

Level of Amputation	Distribution of Levels (N=107)	
	Number	***Percent***
Below Knee	27	25%
Above Knee	46	43%
Upper Limb	17	16%
Unspecified	17	16%
TOTAL	107	100%

UPPER LIMB PROGRAM

As of May 31, 2011 125 persons with upper limb loss have been evaluated by an occupational therapist at PRC. A total of 9 patients have been provided with a prosthesis (s) and received associated training, 5 are awaiting arrival of components and 50 are identified to start the next round of provision. Fitting, fabrication and rehabilitation therapy are done at the HHH/HI PRC. UE sockets have been both ICRC polypropylene or laminated using Otto Bock acrylic resin. Componentry is a combination of Otto Bock from Germany and Hosmer in US. Two HHH local technicians have participated in two-week intensive upper extremity training at Don Bosco P&O program, sponsored by ICRC-Special Fund for the Disabled. These technicians actively participate in the upper limb prostheses service delivery under the direction of expatriate prosthetists.

Challenges include the high costs of components, difficulties with importation and customs, limitations with no Haitian credentialed P&O staff and the ongoing cultural stigmas related to acceptance and utilization of prosthetic limbs.

Figure 1 and 2: Participating in training with new prostheses at the Healing Hands for Haiti/Handicap International Physical Rehabilitation Centre, 16 months post earthquake.

DISCUSSION

These authors were both directly involved with the emergency efforts, responsible for rapid assessment of catastrophic injuries requiring emergent rehabilitation services, including prosthetics. We directly examined patients in 17 field and hospital settings in the three weeks following the earthquake, and interviewed surgical staff at each site. Both authors had been working in Haiti in P&O services and training for 10 years prior to the earthquake, and have had ongoing presence in Haiti since the earthquake, through routine visits and medical advisor to HHH (O'Connell), and onsite as Director of P&O for HHH and Country Director HHH (Ingersoll). We fully concur with the current estimates of earthquake-related amputations, emphasizing that both pre-earthquake and ongoing, there are significant numbers of amputations requiring prosthetic and rehabilitative services. The earthquake has resulted in increased international awareness of the needs and challenges experienced by persons affected by disability in Haiti, and the recognition by the government that a National strategy for both education and training in the rehabilitation professions, and for health services are needed, which includes P&O.

CONCLUSION

Organizations involved in P&O services should work in partnership with the National government, collaborate in data collection and dissemination in order to better inform direction of training and service delivery. In keeping with the ISPO/WHO 2003 statement on the relationship between prosthetics and orthotics services and community based rehabilitation [5], training of community rehabilitation workers should complement formal training programs of Category I-III personnel, not replace them. Therefore formalization of credentialed training programs for Haitian P&O staff is a priority of HHH as well as the ICRC.

Haiti remains a country where life for many remains a day to day challenge. A physical disability impacts ones ability to care for self and family, and with limited national resources in assistive technologies and rehabilitation, an amputation impacts survival. The earthquake in Haiti has resulted in an increase in opportunities for persons with amputation to receive prosthetic rehabilitation. It is imperative that coordinated international efforts continue, to support the development and delivery of credentialed P&O training programs and accessible services throughout the country, not just for earthquake victims, but for all persons affected by disabling conditions.

ACKNOWLEDGEMENTS

The authors would like to thank the thousands of volunteers and organizations from Haiti and around the world whom supported the emergency relief efforts in Haiti following the earthquake, and in particular the staff and volunteers of ICRC-SFD, Healing Hands for Haiti, Handicap International and Team Canada Healing Hands.

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DESIGNING FOR AFFORDABILITY, APPLICATION AND PERFORMANCE: THE INTERNATIONAL TRANS-RADIAL ADJUSTABLE LIMB (ITAL) PROSTHESIS

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ABSTRACT

A multi-disciplined team of engineers, prosthetists, physicians, and amputees has developed a new trans-radial prosthesis currently completing second-year validation testing. The International Trans-radial Adjustable Limb, or ITAL, comprises an innovative variable-compression, variable-geometry interface, a new body-powered prehensor with adjustable pinch force, control harness, and cable. Globally, prosthetic options are increasingly limited by cost and lack of infrastructure required to fit and maintain them. The ITAL was purposefully designed to overcome these barriers, providing a new option for economically disadvantaged amputees in the USA and developing countries. Designed to withstand harsh environments, ITALs are relatively low-cost and restore bimanual capacity to perform strenuous physical labor, enabling amputees to earn a living and be self-reliant.

Packaged in kits, units can be taken directly to amputees and fit or serviced inside approximately one hour using simple hand tools, without requiring amputees visit or be transported to a central facility. Amputees in Jamaica, the USA, and Thailand have shown the ITAL to be an appropriate solution, and now use it for daily activities. Users rate the ITAL's comfort at approximately 75% that of custom-fabricated prostheses but equivalent in utility. Future work aims to further increase comfort while refining the aesthetics to address cultural needs.

INTRODUCTION

Trans-radial amputation largely occurs due to trauma, disease or illness and is not generally the preferred option if healthy restoration and rehabilitation of a functional limb is possible. Incidence rates of upper-extremity amputation in nation states around the world are currently unknown; there are, however, general estimates that more than thirty (30) million people need orthotic and prosthetic services. **Murdoch [1]** estimates that potentially one hundred thousand (100,000) prosthetists would be necessary to meet the developing world's needs using current fitting and fabrication methods. Approximately 80% of the world's population makes less than US$ 2.00 a day, a figure often inferred as directly proportional to the resources of persons with amputation who require P&O services [**2, 3**].

Consequently, the cost of prostheses is a limiting factor in providing access to the vast majority of amputees [4]. In order to provide prosthetic tools to amputees who live in remote, rural, and difficult-to-access areas, the logistics necessitate a design paradigm based on the operating environment, affordability, and performance. A selected approach will only be successful if it correctly accommodates available resources and requirements of users. Clinical experts have urgently encouraged researchers and manufacturers to "strive for developments that ultimately culminate in clinically practical, integrated, and affordable techniques", [**5**] and to develop affordable devices for targeted applications such as farming [6]. Several groups have heeded this call to action; this research and development effort was initiated specifically to create a low-cost, biomechanically appropriate upper-extremity prosthesis for below-elbow amputees.

METHODOLOGY

To initiate the development process, design specifications were obtained from peer-reviewed articles, and a multi-national survey of prosthetists, doctors, and amputees who owned their own devices, and individuals who had access to none. The main measures of UE prosthesis acceptance identified were the comfort, suspension, and aesthetics. Specifically, among many users surveyed, minimum required suspension ranged from 20-25 lbs. Comfort was equally important, with users wanting to be able to perform manual tasks for a minimum of three hours daily. Comfort ranked particularly important for many individuals who depend upon manual labor as their only means of employment. Based upon published literature, it is clear that despite relative simplicity and more limited dexterity, body powered (BP) prosthetic systems remain very popular; the majority of UE amputees either uses them exclusively or keeps them as backup for myoelectric devices [7, 8, 9]. The primary reasons for their popularity are comparatively lower cost, low weight, mechanical robustness, preservation of proprioceptive feedback, and ease of maintenance in comparison to myoelectric devices.

Invariably, even amputees who had no access to devices wanted aesthetic options; function, however, was equally important for many individuals. Passive cooling was

requested by many prosthetists who work with more-active amputees. Prosthetists and doctors working in remote regions repeatedly requested technologies that can be deployed (i.e. fit and adjusted for use) rapidly—within a day—with minimal use of materials, electricity, tools, and the ability to be fit outdoors. In many cases, high corrosion resistance and the ability to withstand constant exposure to dust, dirt, and fluids (including perspiration, brine, and those associated with farming and raising animals) were also specified as performance specifications. Body powered users requested hooks that eliminated the need to use numerous latex bands to provide grip forces sufficient for rigorous tasks.

Collectively, these design and performance specifications describe a simple, relatively inexpensive technology that incorporates flexibility in function and application; especially if alterations for certain users become necessary. Emphasis was placed on balancing cost and functional requirements in a complete prosthesis design which would include all elements necessary for immediate use, including a harness, terminal device (TD), and socket. As an alternative to developing a rigid, fixed-geometry socket, a variable-geometry 'mechanism' was devised to provide most of the requisite suspension. In contrast to a full-contact socket, a patent-pending open-frame "interface design" was developed to permit rapid installation with simple and easily accessed hand tools. A new terminal device was also developed that permits adjustment of pinch force to match task need, while also reducing the physical stresses imposed on users' by existing split-hook that come with the use of multiple bands on a standard voluntary-opening hook TDs. Throughout the design process, materials were selected that are widely available from other industries to reduce the cost of repair and maintenance.

The Interface

A modular two-component interface comprising a "humeral cuff" and forearm adaptor were developed, and each can be manually adjusted with simple hand tools. The humeral cuff was iteratively designed to be comfortable under considerable loads while providing the major proportion of suspension. Figure 1(a) illustrates the cuff location on the humerus. Affixed to the distal humerus, the cuff directly contacts the olecranon process and the medial/lateral condyles—three-point contact to establish stability. Each contact point incorporates a combination of rigid and compliant materials shaped to engage the user's residuum. Condyle contacts were devised based on observations of anatomical structures of biological gripping mechanisms. A mechanism analogous to bone and soft tissue of the finger pads of the distal phalanges was developed, comprising an inner rigid core encapsulated by a combination of elastomers.

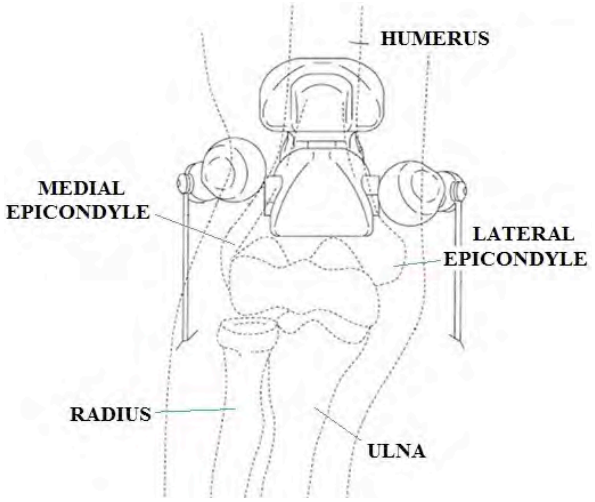

Figure 1: Humeral Cuff positioning on the Distal Humerus

Along with a simple fastening mechanism, this arrangement creates a comfortable mechanical lock surrounding the distal humerus. The epicondyle contact configuration is shown in Figure 2. Correctly adjusted, the cuff effectively minimizes rotation and migration below the condyles, and prevents point loading on the humerus, thereby increasing comfort and consequently also reduces the life of this cuff component.

Figure 2: Epicondyle Contact Configuration

Readily replaceable sleeves minimize the inconvenience of replacing worn parts. In conjunction with the humeral cuff, an open-frame bivalve forearm adaptor design provides additional interface suspension and stability. The humeral cuff is shown suspending 40 lbf (178 N) in Figure 3. Three cuff sizes accommodate the desired range of anthropomorphic dimensions, and can be used on either the left or right arm. Humeral cuffs can be adjusted to fit an amputee in less than one hour using a single Allen-wrench. Together, the humeral cuff and the forearm adaptor constitute a variable compression, variable-geometry open exoskeletal interface.

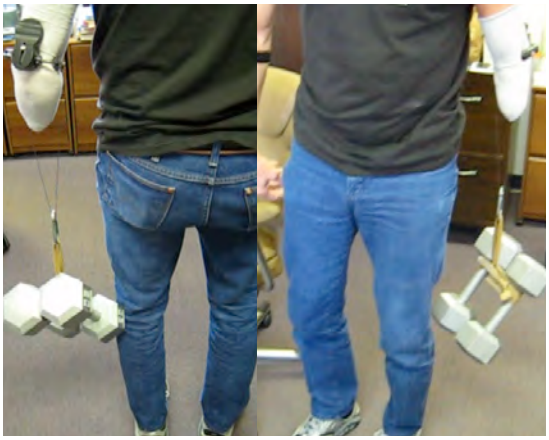

Figure 3: The adjustable humeral cuff Suspension

The Terminal Device

A terminal device was developed that incorporates grip force adjustment. Movement of the band attachment point is accomplished using a patent-pending dual-ramp ratchet mechanism. Moving the band attachment location varies both the initial pre-stretch and the effective tensile force, shown in Figure 4. This allows user-adjustable pinch force to match task need and to conserve muscular exertion. Users select one of six (6) discrete positions by pushing or pulling a carriage tab, making it index to the next available position. This is done using the sound hand or by pressing the tab against nearby objects.

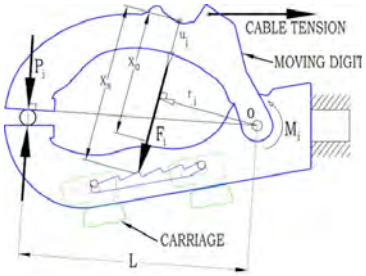

Figure 4: TD schematic and principles of operation

The terminal device uses either a single elastic bungee ring or several inexpensive Size 19 rubber bands, i.e., 3.5 inch x 0.06 inch square (89 mm x 1.5 mm square), commonly sold in bulk packages. Up to 45 bands can be installed to achieve a maximum pinch force of 17 lbf (76 N). The mechanical arrangement of the TD has the bands crossing the grip contact area, providing similar function to the webbing between the thumb and index finger and allowing some shock absorption and added contact on irregular shaped objects.

The TD was also designed with an elastomer contact on the main contact surfaces. This component increases surface friction when utilized during gripping. The "tips" can also be easily replaced when worn. To accommodate different levels of use, cost, durability and weight, the TD was developed for manufacture as a stainless steel, aluminum alloy, and a glass reinforced plastic. Combinations of fiber reinforced plastics are utilized for separate digits of the TD to realize a combination of low cost, strength, flexibility and utility in coastal or marine environments. Gripping contours of the new TD were also developed to limit digit separation when cylindrical grip loads are applied.

The Cable and Harness System

A Figure-of-9 harness was selected for use with the interface and TD for its simplicity. To keep costs relatively low and increase ease of repair, new harness components using fasteners instead of one-shot crimped ball terminals were developed. Bicycle cabling was incorporated into the new cabling configuration as a comparatively inexpensive alternative to current existing cable housing components. Additionally, bicycle cable, conduit and housing, is available worldwide.

RESULTS

This effort resulted in the creation of the International Trans-radial Adjustable Limb® or the ITAL®. This name was chosen because some in remote areas have no language equivalent for the term 'prosthesis'. Among those surveyed, the term 'artificial limb' is already widely accepted and translates in the local language. Volunteer evaluators comprising both new amputees and experienced prosthesis users (i.e. those having used a prosthesis for a minimum of two years) who perform routine robust tasks. A total of 10 amputees were included in the evaluation group. Among experienced evaluators, interface comfort was rated at 70 to 75% of regular custom prostheses. All volunteer evaluators were allowed to keep the ITAL and use it based upon their own preferences and needs. The experienced prosthesis users were willing to use the ITAL for a few hours a week, and for extended periods during recorded tests.

Typically they did not want to give up the comfort of their own interfaces. Comfort and suspension were evaluated doing routine tasks such as changing a car tire, shoveling, and manipulating multiple objects. After refining the design using feedback from the experienced users, volunteer evaluators were identified in Jamaica, Thailand, and Ecuador. During the initial evaluation stages, susceptibility of both the TD and the interface to rust necessitated the redesign of both to incorporate stainless steel and engineering polymers. This became particularly evident during the first week of testing by amputees who lived within a mile of the sea. Six amputees were tracked at different stages over a two-year period. One individual was an experienced prosthesis user, and two individuals received units within 8 months of amputation; the other three were amputees for five years or more and who had not used any device.

Figure 5: ITAL in use for home construction

Figure 5, shows person with the ITAL using it for building construction. The experienced user continues to use the device as a backup unit, for activities including weightlifting, house cleaning, and some sporting activities. Of the two who received the ITAL within 8 months of amputation, one uses the device daily as a farm laborer, and the other uses the device occasionally. The latter of the two was not fully satisfied with the appearance of the ITAL but did use it initially for subsistence-type farming and building wooden structures. Over time he developed considerable more muscle mass in his residuum that caused secondary discomfort. He currently uses the ITAL only for heavy lifting tasks where bimanual capacity is needed.

Contact with another individual has only been by phone as schedule conflicts have prevented directly meeting to discuss usage details. Finally, two have reported satisfaction with the device and they are now able to work in the field more effectively and participate in building additions to their homes. An unanticipated effect of the variable-geometry design, variable tension/compression mechanism is that the evaluators began adjusting the device (both the interface and the TD) in real-time to maximize comfort. For heavy duty or light duty tasks, they sometimes opt to modify the cuff fit to vary the suspension and overall comfort based on anticipated activities. Similarly, they use the dual ramp ratchet mechanism to adjust TD pinch force "on the fly" almost continuously to vary grip based on their current task.

CONCLUSIONS & FUTURE WORK

In its present form, the ITAL is lower in performance than custom-fit BP UE interfaces, but performance was deemed very satisfactory overall. Total manufacturing cost for the basic ITAL unit is less than US$400. Evaluators have expressed a desire and willingness to purchase complete units for personal use, specifically for sporting activities and as backup prostheses. The ITAL is presented as an appropriate alternative prosthetic device for use in both the developed and developing world. Given its high-functionality, low-cost, and ability to be readily distributed, its ultimate benefit-to-drawback ratio may prove significantly higher than available devices. Continuing refinement is focusing on improving the aesthetics of the design to reduce this barrier which remains among many potential users. Prosthetists and amputees involved in this project are currently determining best practices for fitting and adjusting the interface for long-term usage; these recommendations are being captured in a comprehensive fitting instruction set. The ITAL has proven to be an appropriate alternative to standard trans-radial prostheses, especially as an option for the poor and uninsured.

ACKNOWLEDGEMENTS

This work was funded by the United States National Institutes of Health (NIH) through a Small Business Innovation Research (SBIR) grant no. 5R44HD054091. The authors wish to thank Christopher Jones, CPO, of Rebound Prosthetics USA and his staff, Dr Rory Dixon, Sir John Golding Rehabilitation Center Jamaica, The Shan Health Committee Thailand, Jen Lee of the Prosthetics For Life Foundation Ecuador, Allan Oates Prosthetic Solutions South Africa, Veronique Morin, Volunteer Kakuma Refugee Camp Kenya, and the entire staff at Laboratorio Gilete, Columbia. Opinions expressed in this paper are those of the authors and do not necessarily reflect opinions of the sponsors.

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OCCUPATIONAL THERAPY FOR A MULTIPLE LIMB LOSS MILITARY PATIENT, A CASE STUDY

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ABSTRACT

This Severe polytrauma involving multiple limb amputations have unfortunately become more common through the ongoing conflicts in Iraq and Afghanistan. This population of young, motivated, and severely injured patients present unique challenges for therapists and prosthetists. Client centered care is an essential part of positive therapeutic relationships and in achieving functional goals. This case study will give an overview of care by reviewing challenges, barriers, prosthetic training, limitations of prosthetics, and therapeutic use of self to foster best outcomes with this individual.

USING MULTIPLE OUTCOME MEASURES TO DETERMINE SKILL LEVEL IN MYOELECTRIC PROSTHESIS USE

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INTRODUCTION

Most studies on prosthesis usage focus merely on one type of outcome measures, using questionnaires, functional tests, or kinematics. However, a combination of several outcome measures should provide a better picture on prosthesis use (Hill et al. 2009, Lindner et al. 2010, Wright 2009). Using both clinical and more fundamental measures (such as kinematics) would not only provide information about the skill level of a prosthesis user, but would also give insight in the processes from which the level of skill originates. To maximize the insight in the skill level of a prosthesis user the current study gauged a wide range of outcome measures. The aims of this study were 1) to describe prosthetic functioning at different levels of performance; 2) to relate the results of the clinical level to the more fundamental outcome measures; 3) to identify specific parameters in these measures that characterize the level of skill of a user.

METHODS

Six experienced users of a myoelectric forearm prosthesis (mean age 36 years, SD 18 years) volunteered to participate. All participants had a passive wrist rotator.

The Southampton Hand Assessment Procedure (SHAP, Light et al. 2002) was used as a clinical test. SHAP consists of 26 tasks; 12 abstract object tasks—six lightweight and six heavyweight objects—and 14 activities of daily living (ADL) tasks. SHAP evaluates hand functionality and provides an Index of Functionality (a sound hand scores normally between 95 and 100, lower scores reflect decreased hand function). Each task was timed by the participant with help of a timer button.

For the fundamental measures two goal-directed tasks were examined: direct grasping and indirect grasping. Four objects were used in the grasping tasks, three compressible objects, each with a spring of a different resistance, and a solid object. The compressible objects simulated non-rigid objects used in daily life, like a juice carton. All objects were covered with a Velcro strap, which had to be pulled off to simulate manipulation of the object. Movements were recorded with a motion analysis system (Vicon), and a head-mounted eye tracking system (IScan). The participants were instructed to execute each of the tasks as rapidly and as accurately as possible, while trying not to compress the objects.

Because of the individual differences between the participants, the data were analyzed for each participant separately. Time scores of SHAP were transformed to an Index of Functionality score and to z-scores. Mean z-scores were calculated for the lightweight and the heavyweight abstract tasks, and the ADL tasks. The following end point kinematic outcome measures were calculated: reach time, peak velocity of reach, grasp time, plateau time in aperture, termination asynchrony. Compression of the objects was measured to assess grip force control of the prosthetic hand. Two Kruskall-Wallis tests were executed on the dependent variables, and Spearman's Rho Correlation was determined for the mean z-scores of SHAP and the endpoint kinematics. Joint angles were produced with the Plug-in-Gait model of Vicon and the Range of Motion (ROM) was calculated. Gaze behaviour was scored frame by frame with help of Anvil video-annotation software.

RESULTS

All participants scored far beneath the normal Index of Functionality score of 95-100 with SHAP. There was a large difference between the scores, with a highest score of 71, and a lowest score of 17.

The two different grasp tasks influenced mainly the variables of the transport of the hand towards the object, whereas the effect of objects was mainly reflected in the dependent variables of the grasp and object manipulation. Differences between the participants could clearly be noticed in the dependent variables, reflected in differences in time needed to execute the tasks, and the amount of compression of the objects (see Figure 1).

SHAP scores correlated significantly with reach time, peak velocity, and plateau time.

The movement patterns and the Range of Motion for the direct grasping task and the indirect grasping task were slightly different. Although all participants showed overall the same movement patterns in the joint angles, there was much variation in the amount of shoulder abduction between the participants.

Overall, two types of gaze behaviour were noticed. Four of the participants fixated at the object after the start of the trial, and looked at the object most of the time during the trials. The other two participants looked more at the prosthetic hand during execution of the grasping tasks. No differences of gaze behaviour between the different objects were noticed.

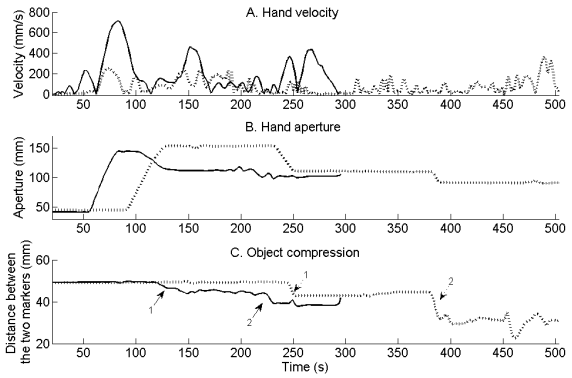

Figure 1: Illustrative example of two participants who performed a direct grasp with a compressible object. The solid line represents the participant who scored the highest on SHAP; the dashed line represents the participant who scored the lowest on SHAP. During the reach of the hand towards the object (A), the hand opened to a maximum aperture, stayed at a plateau for a while, and started to close when the hand was near the object (B). When the object was picked up, two moments of compression could be determined (C). The first compression occurred immediately when the object was picked up (indicated with arrow 1), and the second—farther—compression occurred when the Velcro strip was pulled off (indicated with arrow 2). The difference between the two participants can be clearly noticed in the height of the velocity of the hand during the reach (A), the time needed to execute the task, the length of the plateau in the aperture (B) and the amount of compression of the object (C).

DISCUSSION AND CONCLUSION

By using outcome measures on different levels of description, a good view is provided on the performance of the participants in the various tasks. The results of the different outcome measures were on average in agreement with each other in terms of the level of performance of the participant. The results also supplemented each other as the results of the fundamental outcome measures described in more detail how the participants performed in both the fundamental tasks and the clinical task. The participants that scored higher on the SHAP showed overall better performance in the fundamental outcome measures: they had smaller movement times, more gaze behaviour towards the object than towards the prosthetic hand, and less compression of the objects. This indicates that SHAP has a good discriminative ability for the skill level of the prosthesis user. Moreover, the correlation between SHAP score and the fundamental outcome measures reach time and plateau time in the aperture indicate that these variables are specific discriminative parameters that underlie the level of skill of a prosthesis user. This is very useful in rehabilitation, as one can specifically focus on the discriminative parameters on which an individual scores low. This could enhance the overall skill level of an individual.

ACKNOWLEDGEMENTS

We thank the Anna Fonds and the ISPO-NL for the financial support of this study.

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OUTCOME MEASURE RESULTS OF PATTERN RECOGNITION CONTROL OF A MULTIFUNCTION HAND-WRIST SYSTEM: A CASE STUDY

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INTRODUCTION

Pattern recognition (PR) has been described as a method of controlling a larger number of prosthetic arm movements than those that are possible with currently available commercial myoelectric control devices.[1] PR has also been used by individuals with higher levels of amputation who have had targeted reinnervation to control advanced prosthetic components.[2]

Previous testing was performed on five individuals with transradial amputations using a virtual reality system with 10 wrist/hand movements. The performance of the participants using the residual limb was compared to their performance using the intact arm.[3] Performance metrics included motion selection time, motion completion time, and motion completion ("success") rate. Classification accuracies with the residual limb (approximately 79%±11%) were not as high as with the intact arm (94%±3%). When only one hand movement was tested, residual limb classification accuracy increased to 93%±4%.

Work to date has now shown that PR can also be used for transradial amputees (without targeted reinnervation) to control a physical device: a multifunction hand-wrist system with seven degrees of freedom (DOFs), including wrist pronation and supination, wrist flexion and extension, hand open, lateral/key grip, and opposition/pinch grip (Figure 1).

BACKGROUND & CONFIGURATION

The subject was a 62-year-old male who sustained a transradial amputation approximately 25 years ago. We compared his ability to control his existing two-site myoelectric prosthesis with his ability to control a multifunction hand-wrist system with PR (Figure 1).

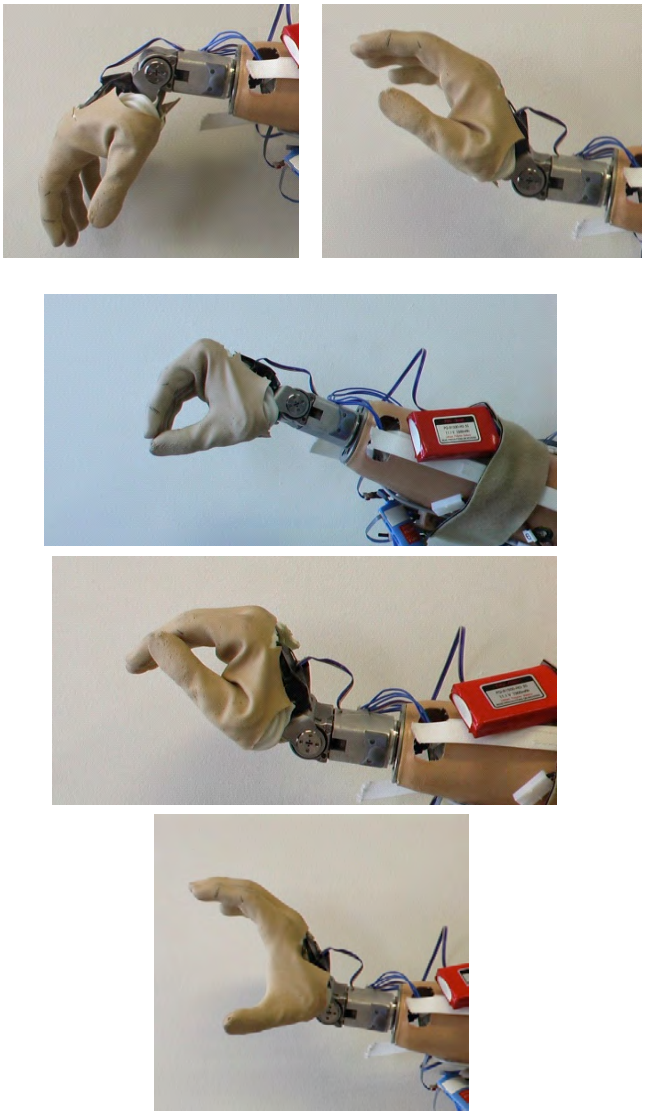

Figure 1: Range of motion of the multifunction hand-wrist system. Device at maximum range of motion of wrist flexion (top left) and extension (top right), pinch and lateral/key grip (center), and hand open (bottom). Wrist rotation moves through 360 degrees.

The existing (or *home*) system was a two-site myoelectric prosthesis with a single DOF hand (Otto Bock DMC) and a nonpowered quick-disconnect wrist. It was self-suspending with a pelite liner. The subject had worn his myoelectric prosthesis for many years. He used the device for specific tasks, especially biking and weight-lifting, but did not use it every day.

As stated above, the multifunction device had seven DOFs and was controlled using myoelectric PR. It was only used within the laboratory. It was fit with a socket and a gel liner with lanyard suspension and six modified Otto Bock electrodes (analog filtering was removed to record appropriate signals for PR). Myoelectric signal processing was done in real time. Electrode signals were sent to the computer via a Bluetooth connection; using PR, the software program then determined the appropriate movement class and wirelessly returned motor command signals to the hand. The PR classifier was retrained at the beginning of each visit. During pattern classifier training, 3 to 6 seconds of data were collected as each movement was performed and were used to create the pattern classifier.

METHODS

Initial Training

Initial training with the multifunction PR system was done using a virtual environment. The experimental socket was fabricated during this time. The subject worked with the arm over the course of a year as software and hardware were developed, with visits occurring approximately one to two times per month for 2 hours at a time. During these visits, tasks performed included manipulating many objects of different sizes, weights, and fragility Items from the various testing kits—including the Southampton Hand Assessment Protocol (SHAP), Box and Block Test, and Clothespin Relocation Test—were used during training to practice various functions. Once the subject was comfortable with the function and control of the device, the pattern classifier was created with as little as 3 seconds of data for each DOF.

Figure 2: Subject moving blocks during training

Since the subject was an established user of his home device, therapy with this device was not performed as part of these experiments.

Testing

Testing was done with both the multifunction system and the home system. Data were collected from the two devices on different days. The outcome measures and functional tests included the SHAP[4], Jebsen-Taylor Hand Function Test[5], Box and Block Test[6], UNB Test of Prosthetics Function[7] (with self-selected, age-appropriate tasks), Assessment of Capacity for Myoelectric Control (ACMC)[8], Clothespin Relocation Test, and a cup-stacking test. For the Clothespin Relocation Test, the subject was required to move three clothespins from a lower horizontal bar to a higher vertical bar using the hand functions and wrist rotation. For the cup-stacking test, the subject removed cups from an inverted stack and then placed six inverted cups into a pyramid configuration with a seventh cup placed right-side-up on the top of the pyramid, thus using all available DOFs.

RESULTS

Results of the various tests are shown in Table 1. Better scores are highlighted in gray. The subject performed better using his home device on all of the various measures except for the Box and Block Test and the UNB test; however, not all of the prosthetic DOFs were utilized during UNB testing with the multifunction system.

Table 1: Results of Outcome Measures.

Outcome Measure	Device	
	Multifunction	***Home***
SHAP: Index of Function Score	47	66
Jebsen-Taylor Total Score (sec)	325	224
Box and Block Test (num. of blocks)	18	11
UNB		
Total Time	502.21	937.32
Total Spontaneity Score	40/40	40/40
Total Skill Score	31/40	35/40
ACMC	0.55	1.24
Clothespin: time (sec)	22	12.75
Pyramid Cup Stacking	63	46.62

DISCUSSION

The subject was able to complete all testing tasks using the additional DOFs of the multifunction system. Some scores were close between the two systems, but in general, use of the additional DOFs of the multifunction system came at the cost of increased time. For example, compensating for a lack of wrist rotation by using shoulder abduction was faster than using the wrist rotator in the multifunction system.

As a result, the home device scored more favorably on all timed tests except the Box and Block and UNB. The ability to position the hand into flexion may have improved the score on the Box and Block Test. The multifunction prosthesis demonstrated a faster time on UNB tasks. This was because the subject had difficulty opening the hand of his two-site system. The skill level was higher with this device, however, because when using the multifunction system, we found that despite asking him to do the task in what we thought would be an appropriate way (as the nondominant hand), he often was insistent that he would "show us what it could do" and therefore did the tasks in a less natural way, using the prosthetic hand as the dominant extremity. For example, when removing money from a wallet, he stabilized the wallet with the intact hand and removed the money with the prosthesis; when using a dustpan, he held the whisk broom with the prosthesis; and, when tearing tinfoil, he held the box with the intact hand and tore the foil with the prosthesis.

One of the reasons the subject did not perform as well with the multifunction system in the standardized tests is that he was less familiar with it. Our next goal is to have a home trial to allow the subject to become better at controlling and incorporating the device into daily tasks as an assistive device. As previously mentioned, for tasks where he was allowed to choose how the device was used (UNB and ACMC), he often inappropriately chose to use the prosthesis as the dominant hand.

FUTURE WORK

Although we have shown that it is possible to control a two DOF hand and two DOF wrist with PR, there are still many factors that need to be resolved before this system can be viable as a home system. For example, the processing that is done on an external computer needs to be transferred to an embedded system. Additionally, a system that detects when an electrode stops providing dependable signals (e.g.. loses contact with the skin) may provide more reliable control. Also, six electrodes can be difficult to integrate into a transradial system due to socket size constraints and the difficulty in keeping six electrodes in contact with the skin through all movements of the residual limb. Efforts are underway to improve the socket function and comfort for home trials

Once hardware issues have been resolved, home trials are planned. These trials are expected to begin this year.

ACKNOWLEDGEMENTS

We would like to acknowledge Robert Parks and Dat Tran for software support, and Aimee Schultz and Ann Barlow for editorial support.

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A PRELIMINARY STUDY OF GAZE BEHAVIOUR AND UPPER LIMB KINEMATICS IN TRANS-RADIAL USERS

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INTRODUCTION

Tools to evaluate the effectiveness of upper limb prostheses generally measure user's performance on functional tasks or use questionnaires to examine the effect of amputation and prosthesis use on everyday life activities [1]. There have been few studies describing the characteristic changes in motor behaviour associated with learning to use a prosthesis [2, 3]. Such studies potentially provide useful insight into the characteristics that are reflective of skill acquisition and hence may lead to the development of improved outcome measures.

Nevertheless, most of these studies only investigated planar pointing tasks, in which no active involvement of the hand is required and relatively small differences in performance between amputees and anatomically intact controls were observed. In contrast, previous studies of activities of daily living (ADL) performance have shown clear differences in joint kinematics and task completion time between healthy subjects and amputees [4]. Moreover, despite the widespread agreement regarding the role of the vision in prosthetic use [3, 5-8], and the extensive literature on the role of vision in learning to use a tool (e.g. [9, 10]) and in the performance of ADLs [11], gaze behaviour in upper limb prosthesis users has received limited attention [3].

In this study, we evaluated the changes to kinematics and gaze behaviour associated with learning to use a myoelectric prosthesis for the performance of an ADL task. We chose to study anatomically intact subjects to allow for comparison of task performance with the prosthesis against performance using the anatomical upper limb. The study firstly aimed to describe characteristic factors which differentiate upper limb task performance with the anatomical hand from performance with a myoelectric prosthesis. The second aim was to identify those factors which changed with skill acquisition while learning to use the prosthesis. Due to space limitations, in this paper we describe the methods and present detailed results from the gaze behaviour part of the study. However, a full set of results, including the kinematics, will be presented at the conference.

METHODS

The study was approved by the University of Salford Research Ethics committee. Following written consent, five anatomically intact, right-handed individuals, (3 males and 2 females) with a mean age of 30 years (ranged from 26-41) were recruited. All subjects were in good physical condition and had within-normal visual acuity. All data were gathered in the Movement Science Laboratory at the University of Salford, Salford, UK.

The experimental setup is discussed in more detail in [12] and hence only brief details are provided here. The study was a cross-over design (Table 1). Participants' gaze behaviour and upper limb kinematics during the performance of an ADL task were evaluated twice in separate sessions (E1 and E2) which formed a baseline phase. Following this they were fitted with a myoelectric prosthesis simulator (see [12]). Subjects were then evaluated 3 further times over the course of approximately 2 weeks, when performing the task with their prosthesis (E3-E5). We also provided 6 separate practice sessions (P). During each of these sessions, subjects performed the Southampton Hand Assessment Procedure (SHAP) once [13]. The SHAP sessions were performed on different days to the evaluation sessions, to avoid fatigue. The SHAP test not only provided an opportunity for participants to practice, but also allowed for an evaluation of hand function over the course of the study.

The ADL task carried out in each of the evaluation (E) sessions involved reaching for a carton and pouring water from the carton into a glass, then replacing the carton on to the table. The task was challenging to perform with the prosthesis and had a cost (water spillage) associated with poor performance. At each evaluation session (E) subjects repeated the task 12 times and the first 10 repeats in which good data were collected were considered for analysis.

At the start of each evaluation session the subject was seated upright in a chair with his/her back resting on the chair back, the upper arm in a neutral position and both hands resting comfortably on the table. The location of the hands when rested on the table were then marked (termed reference

positions) to serve as the start points for each repetition of the task. At the start of each attempt to complete the task, the subject was instructed to initiate the movement from the reference position and to return to the reference position at the end.

Table 1: Experimental protocol

	Phases	Events	Data collection
0	Baseline phase (Anatomical hand)	E1: Task performance (12 repeats)	Kinematics Gaze
		P1: Practice session	Hand functionality
		E2: Task performance (12 repeats)	Kinematics Gaze
		P2: Practice session	Hand functionality
~2 weeks	Training phase (Prosthesis)	E3: Task Performance (12 repeats)	Kinematics Gaze
		P3: Practice session	Hand functionality
		P4: Practice session	Hand functionality
		E4: Task performance (12 repeats)	Kinematics Gaze
		P5: Practice session	Hand functionality
		P6: Practice session	Hand functionality
		E5: Task performance (12 repeats)	Kinematics Gaze

Prior to starting each attempt at the task, the subject was instructed to gaze at a marked point (termed the gaze reference point or GRP) placed in the middle of the table. The GRP was a visual start and end point for all subjects throughout the testing. During task completion, subjects were free to move their eyes as they wish. Furthermore, no constraint on head movement was applied during the task performance. At the end of task completion, the subject was instructed to return their gaze to the GRP. When the prosthesis was used, the table was moved forward relative to the chair to accommodate the extra-length of the prosthesis.

Kinematics instrumentation

Kinematic data were calculated from the positions of reflective markers located on the subject's upper body, collected using the Vicon 612® motion capture system (Vicon Motion Systems, Los Angles, USA) (Figure 1). Marker positions were sampled at 100 Hz. Further details on the analysis and results from this part of the study will be presented at the conference.

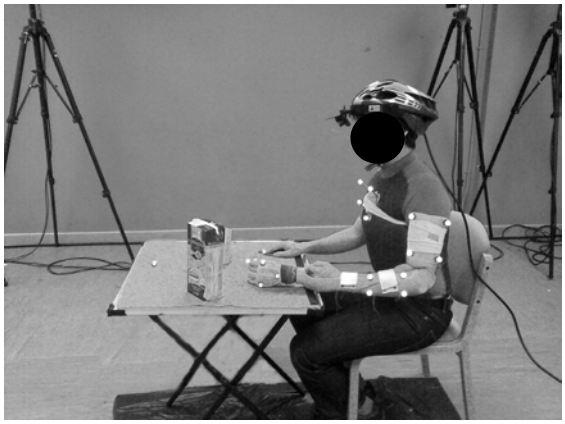

Figure 1: Experimental setup

Gaze data

Gaze data while performing the task were captured using a head-mounted Eye-Tracking system, iView X™ HED 2 (SenseMotoric Instruments GmbH, Tellow, Germany). This system is a head-mounted tool, which continuously tracks the movement of the pupil and projects the location of gaze in a scene video, collected from a head-mounted camera, thus allowing the overlay of gaze position on the scene video to be invariant of head movements. The method for gaze data analysis has been discussed elsewhere in the conference proceedings [12]. In summary, we divided the scene ahead into discrete areas of interest (AOIs) that allowed the pattern of gaze fixations to be described. In each recorded trial, and for each phase, the duration of fixations at each of the AOIs were identified and normalised by the phase duration. For each subject and for each of testing session, the normalised fixation durations of each of AOIs and that for all coded trials in each phase were summed and averaged by the number of trials (n= 10). Then for each of testing sessions, an average of the normalised averaged fixation duration was calculated for all subjects

Functionality scores

The SHAP test produces a functionality profile, based on the time taken to complete each of the 26 tasks [13]. From this profile, an overall functionality score is calculated, using the web-based software produced by the developers of the evaluation tool (http://shap.ecs.soton.ac.uk/entry.php)

RESULTS

In all the graphs, for ease of interpretation, and where appropriate, a dashed line is used to separate anatomical hand's data from prosthetic hand data. Error bar indicates 1 standard deviation (STD) in all cases.

Hand function (SHAP) scores and task completion time

Table 2 shows the mean SHAP scores of all subjects gathered during the practice sessions and time to complete the manual tasks (pouring water from a carton into a glass) across the evaluation sessions.

Table 2: Mean (STD) SHAP scores in practice sessions (P) and time to complete the manual task across evaluation sessions (E).

	E1	P1	E2	P2	E3	P3	P4	E4	P5	P6	E5
SHAP		94 (1)		96.4 (2.1)		36.8 (6.7)	51 (3.3)		60 (6.4)	67.4 (4.5)	
Task duration	11.2 (2.8)		10.2 (1.8)		17.6 (2.7)			16.0 (2.1)			14.8 (1.6)

Gaze data

In this paper we only present gaze data from sessions E1,E3 and E5. The normalised average total fixation durations at every AOI for all subjects across the key sessions (E1, E3 and E5) are illustrated in figure 2.

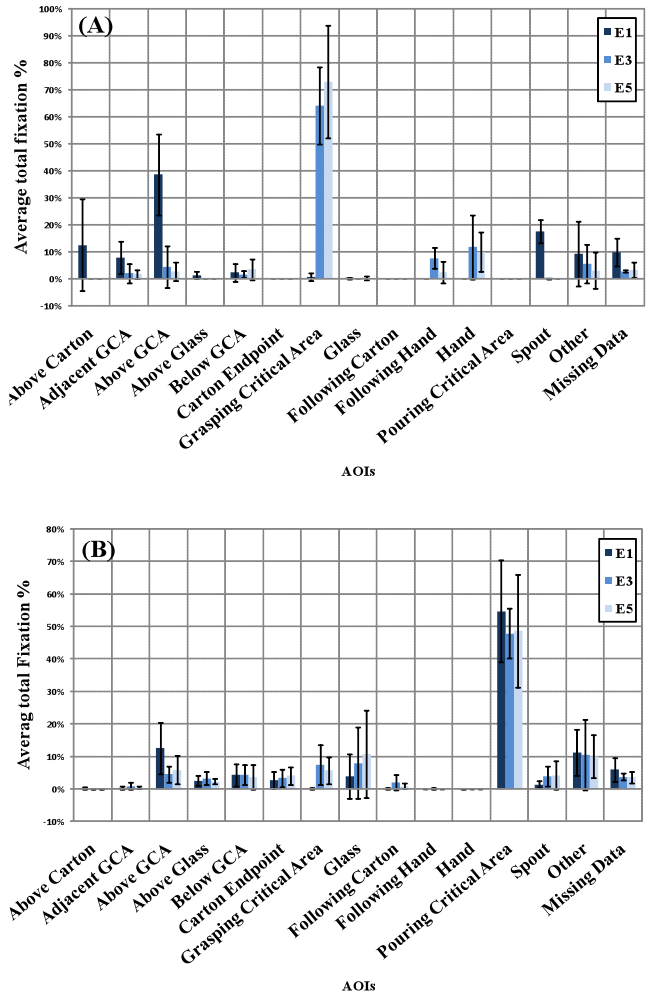

Figure 2: The normalised average total fixation duration at AOIs across sessions E1, E3 and E5 for all subjects, (A) during reaching phase and (B) during manipulation.

DISCUSSION

SHAP and task completion time

The results of the SHAP tests indicate a clear effect of the introduction of the prosthesis on functionality. SHAP scores declined dramatically from around 95 in the baseline (session P1 and P2) to 36 on the first session with a prosthesis (session P3). The effect of practice is evident by the steady increase in SHAP scores witTime to complete the manual task in the evaluation sessions also increased significantly on introducing the prosthesis (Table 2). However with practice, and in keeping with SHAP results, a steady decrease in time to complete the task was observed.

Gaze data

With regard to the gaze results, for ease of discussion, we will focus only on the small number of AOIs that either showed major changes in the duration of fixation between baseline and session E3, or showed significant changes with training (large difference between E3 and E5).

Reaching phase

As can be seen in Figure 2A, there were major and relatively invariant differences in gaze behaviour between anatomical and prosthetic reaching. In line with previous research [11] , during reaching with the anatomical hand subjects did not generally focus either on the hand or its associated AOI (following the hand). Instead, while reaching subjects tended to fixate their gaze mostly at the areas of relevance to the subsequent action (look ahead fixations [14]) (68% of total fixation), notably at "Above Grasp Critical Area", "Spout" and/or "Above Carton" which allows planning to the action ahead in time.

In stark contrast to reaching with the anatomical hand, prosthetic reaching was mostly initiated with gaze fixation at the "Grasping Critical Area" (GCA) (64% of fixations) and, in some subjects, occasional fixation at the prosthetic hand (Figure 2A). While reaching, subjects most often pursued the prosthetic hand and/or flickered between the Hand and GCA. The attendance to the GCA may indicate concern with the hand-carton interaction aiming to correctly and securely grip the carton. Attention at the "Hand", and "Following Hand" AOIs may be associated with concern regarding the hand configuration and location. Attention to these areas (GCA, Hand, and Following Hand) largely precluded the subjects from planning for the manipulation phase.

It appeared that with practice, the duration of the fixation at GCA during reaching to grasp increased slightly, probably as a result of a shorter fixation on the hand area. Such a change in the gaze behaviour may reflect the ability the subjects to incorporate the prosthesis in the internal model of the arm.

Manipulation phase

In contrast to the reaching phase, changes in gaze behaviour during the manipulation phase between baseline and prosthetic sessions were not as clearly differentiated (Figure 2B). Nevertheless, unlike when using the anatomical hand, using the prosthesis required noticeable attention to the GCA during the manipulation phase (8% of total fixation). This may reflect the lack of the reliable feedback from prosthesis regarding the hand state.

While using the prosthesis, it is noticeable that subjects fixated more on the "Glass" (from 7% to 11% of total fixation) and less at Pouring Critical Area (from 55% to 48%) which is probably due to the poor sensory feedback via the prosthesis to estimate the remaining amount of water in the carton.

CONCLUSIONS

Despite the dramatic improvement in prosthetic technology, the extent to which amputees make use of their prosthesis in everyday life to perform functional tasks still appears to be low. This study has shown that gaze behaviours clearly change when compared with those during performance of an everyday task with the anatomical limb. Smaller, but still noticeable changes in gaze behaviour were observed with learning to use the prosthesis. A future study in an upper limb amputee population will investigate whether this characteristic may help to explain observed differences in prosthetic usage.

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MOTION ANALYSIS TO MEASURE OUTCOMES FOLLOWING TARGETED MUSCLE REINNERVATION SURGERY

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INTRODUCTION

The 'Box and Blocks' task (BB) is a component specific task of performance which tests the gross manual dexterity of the upper limb (UL)[1] and is a widely used outcome measure for UL function in several populations.[2-4] The BB task has also been used to examine UL prosthetic function following amputation, and specifically used to examine myoelectric control after Targeted Muscle Reinnervation surgery (TMR).[5-7] The BB task measures speed and quantity with respect to the number of blocks moved in a specific time period,[1] regardless of quality of movement or compensatory strategies used by the individual to accomplish the task.

However, restoring "normal" UL function after amputation relies not only on quantitative performance but also on qualitative observation of the smoothness of pattern of motion and the ability to target and control excursion of the grasp.[8-9] In addition, it is important for prosthetic users to minimize compensatory motions of the body when adapting to the limitations of a prosthetic device. With advances in UL management, such as TMR surgery, the need to accurately quantify the advantages of such procedures is even more essential, as subjective interpretation by the patient and observations on the quality of movement can be more impressive than timed tasks or traditional outcome measures of upper limb function.[10] With a specific repetitive task such as the BB task, the arc and smoothness of motion of the prosthetic limb can be repeatedly observed and recorded; and is amenable to motion capture for the purpose of quantifying this component of performance. Our goal was to establish a method to quantify this improved quality of movement of TMR patients using myoelectric prostheses.

The purpose of this report is to describe a method of quantitative motion analysis, in combination with a modification of the BB task, used to quantify the observed improvements in compensatory movements and control in a subject pre- and post-TMR surgery.

METHODS

Subject

The subject was a 28-year-old male with traumatic left transhumeral amputation on July 4, 2006. The subject was initially fit with conventional body powered prosthesis with mechanical voluntary opening terminal device, and was a successful daily prosthetic user. The subject underwent transhumeral TMR surgery[2] 20 months post amputation and 8 months post surgery, he was fit with a TMR control myoelectric prosthesis. The motion analysis testing of the subject occurred prior to TMR surgery using the conventional prosthesis, and 3 months after fitting with the myoelectric TMR prosthesis.

Motion Analysis

A total of 6 markers were applied to the subject, including sternum, C7, acromion (bilaterally), lateral elbow hinge, and wrist. There were three markers placed on the box and divider to identify the location of box and blocks in the virtual labspace as well as the subject's location relative to the box (Figure 1). Motion was captured using eight Motion Analysis Corporation cameras with a sample frequency of 60 hertz (Hz).

Modification to the BB task

Rather than a random assignment of blocks, the placement and order of blocks in the tray to be moved was standardized to 16 blocks placed in 4 rows. The subject was instructed to proceed from lower left corner block, across the row then proceed to the next row. It was felt that this set up would require specific targeting of the terminal device and demand for consistent activation and arc of movement that would require precise control and be amenable to motion analysis. In addition to the kinematics of the prosthetic and trunk motion, the time to complete moving all 16 blocks was recorded or the number of blocks moved within 1 minute, whichever the subject accomplished first.

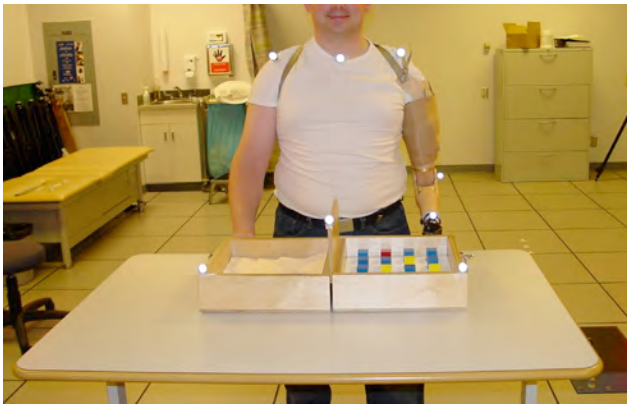

Figure 1: Marker placement and set up for motion analysis

RESULTS

The subject moved fewer blocks with the myoelectric prosthesis than the conventional prosthesis (Table 1). With the modified task, it took 56 seconds to move 16 blocks with the myoelectric prosthesis, and 29 seconds with the conventional. However, this faster time with the conventional prosthesis was associated with the use of a locked elbow versus normal elbow motion demonstrated with the myoelectric prosthesis. In addition, the locked elbow prosthesis required excessive trunk compensatory motion compared to the trunk motion recorded with the myoelectric prosthesis, which was close to motion of normal subjects.

Table 1: Results of standard and modified tasks for BB

	Prothesis Used	
Task	*Conventional*	*Myoelectric*
Standard BB Task (# of Blocks moved in 60 sec)	49	29
Modified BB Task (Time (sec) to move 16 Blocks)	20	56

DISCUSSION

In the case study presented, one of the more striking findings is that while the subject clearly had slower performance with the myoelectric, the movement control was better and less compensatory adjustments were required to perform the task. The subject in this case report also reported increased naturalness of movement and less mental effort to operate the myoelectric prosthesis. Although the results clearly demonstrate that the speed of block movement was much slower with the myoelectric than the conventional prosthesis (less than half), the prosthetic and trunk motion recorded using the myoelectric device was much closer to that of a normal subject. The speed difference may be a factor of the subject being an experienced conventional prosthetic user for 18 months; versus the newly fitted myoelectric prosthesis 3 months post TMR surgery.

As this was a retrospective case study, a minimal marker set was used to answer a specific clinical question and to document overall quality motion following surgery. It is suggested that for future studies, other compensatory motions such as shoulder motion and trunk rotation be included in the kinematic analysis with a larger standard marker set. The modification to the BB task is a simple adjustment to allow quantitative motion analysis to capture quality of motion. Future studies will focus on collective normative data for the modified task to allow larger population comparisons.

CONCLUSION

The current study presents a novel approach to quantifying quality of motion with a modification of the BB task and motion capture in a transhumeral prosthetic user pre/post TMR surgery. With further study, the modified BB task with motion capture has the potential be a useful standardized outcome measure for a variety of upper limb impairments due to its ability to quantify motion patterns of the upper limb as well as compensatory body movements.

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USING MOTION ANALYSIS TO AUGMENT UPPER-LIMB PROSTHETICS OUTCOME MEASURES

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INTRODUCTION

As upper-limb prosthetic systems have become more sophisticated in terms of actuation and control, greater importance is being placed on producing movements of both the prosthesis and the user that have a more physiologically normative appearance. Reducing the degree or range of compensatory motions and improving the trajectory traversed in the movement of multi-joint prostheses are examples of design objectives to bring about better dynamic appearance. Standardized outcome measures in upper-limb prosthetics research and assessment generally do not provide objective information on the quality of movements performed by the prosthesis user. Without this type of information, it is difficult to evaluate different prosthesis designs for their impact on kinematics.

Early efforts to study the kinematics of human movement and apply the results to the design or understanding of arm prostheses were based on cinematography [1], [2]. However, digitizing sequences of individual movie frames to extract joint angles and trajectories was exacting and time consuming work and proved sufficiently daunting to limit its application. The replacement of film with magnetic videotape provided almost instantaneous access to the recorded image, but had little impact on the process of digitizing the individual images to extract the movement data. It was not until the advent of automatic motion capture with multi-camera systems recording the position of passive or active markers on the body that kinematic studies became widely attractive.

The point has been reached where motion data is, relatively speaking, easy to obtain and the problem of acquiring the data has been supplanted by questions of how can kinematic studies help our understanding, how much confidence can we have that the acquired data represents what we think it represents, and how best can the data be represented to provide insight into the important features of the movement under study.

ROLE OF KINEMATIC ANALYSIS

In 2009, the American Academy of Orthotists and Prosthetists convened a State-of-the-Science Conference on Upper Limb Prosthetic Outcome Measures [3]. The conference participants reviewed a variety of measures and assessment tools in use at the time and ranked the tools with regard to their methodological strength and field of application. None of the recommended outcome measures included motion-capture kinematic analysis. Furthermore, none of the measures ranked as "emerging", "promising", or "potential" incorporated motion-capture kinematic analysis. What, therefore, is the role for kinematic analysis?

Motion-capture kinematic analysis is not an outcome measure. It can be a measurement that when used in conjunction with a standardized outcome measure helps to clarify and enhance the understanding of results obtained from an outcome measure. Kinematic analysis provides objective information about the specific actions performed by an individual in carrying out a task.

Although kinematic analysis has the potential to aid our understanding of upper-limb prosthesis use and utilization, it is not without its caveats. Motion capture systems, widely used, require markers that are placed on the body. From the location of these markers, it is possible to approximate joint centers and segment lengths.

Ideally, markers should be anchored to the skeleton, but that is rarely feasible. Instead, markers are attached by adhesive to soft tissue overlying palpable skeletal structures. Skin movement that changes the location of the marker can alter the apparent location of a joint axis and/or change the apparent length of a limb segment.

There are several standardized marker sets that define where on the body markers should be placed for different kinematic analyses [4]. There are not, however, standardized sets that include prostheses, and investigators or clinicians

are left to their own judgment as to where to place markers on prosthetic devices. Not only could different marker placements potentially alter results and confound inter-study comparisons, movement of the prosthesis on the residual limb could affect the apparent location of markers placed on a prosthesis. Pistoning or angular displacement of the socket, especially near the extremes of joint motion, could create a pseudo-arthrosis, reducing the accuracy of the location of markers on the prosthesis with respect to markers on the body. For example, one study involving a subject with a transhumeral amputation who used a locking mechanical elbow was found to have a range of about 13° of elbow flexion when the elbow was locked [5]. The authors of that study attributed this finding to possible relative motion between the socket and the residual limb.

In addition to artifacts that affect apparent marker location, the subject may also behave differently while wearing markers and produce movements that are not representative of how the subject would move without the markers. Markers placed on the upper limb are readily visible. Their presence may distract the user or make the user guarded so as not to dislodge or bump a marker while performing a task.

REPRESENTING MOTION DATA

Motion-capture data—the sequential series of joint angles and limb and body segment positions sampled during an activity—can be further processed and reduced in various ways. One method is to calculate the angular range of motion during the activity for each joint of interest [5], [6], [7], [8]. The ranges, along with their minimum and maximum values, could be compared between subjects with intact limbs and subjects with impaired or prosthetic limbs doing the same task, or between subject using different types of prostheses. Differences in angular range of motion may reveal compensatory actions or effects of varying components or socket designs.

Ratios of angular ranges have been used to define asymmetry between right and left limbs used in a symmetrical bimanual task, such as picking up a box [7]. Ratios of angular ranges have also been used to highlight limitations of motion and compensation [8].

Data obtained during cyclic or repeated activities can be normalized with respect to where they occur in the cycle or at what percentage of time over the course of the activity they occur. [5], [7], [8], [9]. Normalization eliminates absolute time, enabling averaging of repeated data that occurred during shorter or longer periods for an individual subject and comparison of averaged data between subjects. Normalization can be used to compare angular profiles, trajectories, velocity profiles, and relative timing of coupled or uncoupled actions.

CONCLUSIONS

Kinematic data obtained from marker-based motion-capture systems has been used to reveal and highlight differences between actions performed by persons with intact upper limbs and persons with impaired upper limbs or with persons using arm prostheses. Although not an outcome measure in itself, kinematic analysis might be a powerful complement to standardized outcome measures, providing details about the movements used during functional tasks that individual outcome measures alone cannot convey. Attention to methodology and awareness of errors that prosthetic systems might introduce into kinematic analysis are important aspects for successful application.

ACKNOWLEDGEMENT

This work was funded by the National Institute on Disability and Rehabilitation Research (NIDRR) of the Department of Education under grant number H133E080009 and is supported in part by the Jesse Brown VA Medical Center through use of the Jesse Brown VAMC Motion Analysis Research Lab. The opinions contained in this publication are those of the grantee and do not necessarily reflect those of the Department of Education nor do they necessarily reflect the opinions of the Department of Veterans Affairs.

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USE OF A DYNAMIC LOAD STRAP IN ADJUSTABLE ANATOMICAL SUSPENSION FOR TRANSRADIAL AMPUTATIONS

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ABSTRACT

Background: Although dynamic load straps have been used in prosthetics and orthotics since the 1950s, to our knowledge, their use for suspension control in an upper extremity prosthesis has not been previously reported.

Purpose: We will present a case study that used a dynamic load strap as the posterior strut of a self suspending transradial prosthesis.

Method: This work was part of the VA study to optimize the DEKA arm. The subject was a 62 year old male. His right arm and leg had been amputated 43 years prior in a traumatic accident. He used the DEKA Arm with the dynamic load strap within the laboratory setting for 13 visits, over three weeks. The socket was a modification of the high fidelity design.

Results: The dynamic load strap served as an adjustable posterior strut, allowing the user to adjust the amount of suspension for comfort and activities. Use of the dynamic load strap eased the donning and doffing process, and allowed for preservation of pronation and supination movement.

Conclusions/Implications: The dynamic load strap was an effective option for use in suspension the transradial prosthesis. Dynamic load straps might also be useful within harnessing for sockets at other levels of amputation.

INTRODUCTION

The abandonment rate of upper extremity prostheses is high. [1-6] Many studies cite the socket as a contributor to abandonment. [4,6,7] There are several types of sockets used for transradial prostheses.[8] The conventional socket uses flexible or rigid hinges and a triceps cuff for suspension. The use of flexible hinges allowed for pronation and supination. Self suspending designs including the Munster and Northwestern style sockets were developed. More recently, the anatomically contoured socket was developed[9]. This single case study we describe was part of a larger 4 site study of the DEKA Arm System. As of March 2011, a total of 26 subjects had been enrolled. The Generation 2 DEKA Arm is a modular arm which can be provided at the transradial, transhumeral, and shoulder disarticulation/ scapulothoracic amputation (shoulder configuration) level. The subject we describe was a transradial amputee who used the radial configuration of the DEKA Arm. The socket style used in this study was the "high fidelity" style developed by Randall Alley subsequently modified for suspension with a Dynamic Load Strap.

Dynamic Load Strap is made up of a fiber braid with an interior pneumatic bladder. As the bladder inflates, the strap shortens. This provides an adjustable mechanism which does not require any buckles or snaps. One advantage of use of a flexible posterior suspension is the preservation of any native pronation and supination available.

METHODS

This work represents a single case of a transradial amputee who used a dynamic load strap as a posterior strut. The subject was a 62 year old white male Veteran. He had a traumatic distal third transradial amputation 43 years ago. He was a proficient user of a body powered system, although he rarely chose to wear it. He owned a myoelectric, but had rejected it. He had a sensitive pressure point near the distal end of the residuum near the distal radius. The initial check socket was rigid, and the dynamic load strap was added to the final socket.

Figure 1: Proximal and Posterior view Socket with Dynamic Load Strap. The posterior strut was comprised of the dynamic load strap. The attachment nozzle for the airbladder is shown in the lower right.

The subject was casted for a prosthesis. He was then seen for a diagnostic fitting with a rigid socket. In the final fitting, the rigid posterior strut was removed and a dynamic load strap was used in its place. The completed socket is shown is Figure 1. A comparison of the diagnostic and final socket is shown in figure two.

Figure 2: Comparison of the rigid diagnostic socket and final socket with the dynamic load strap. The anterior and posterior openings were maintained, the posterior strut was replaced with the dynamic load strap, and the medial and lateral opening were removed to improve capture of native pronation and supination.

The subject used the DEKA Arm for 10 training visits lasting 2 hours each and 3 testing visits lasting approximately 3 hours each. We estimate that he had about 30 hours of wear time over the course of the study. The completed arm with DEKA hand is shown in figure 3.

RESULTS

The dynamic load strap comprised the posterior strut, and provided suspension for the prosthesis. The dynamic load strap increased comfort, and was easier to don and doff than previous versions of the socket. In addition, the dynamic load strap allowed some native pronation/supination. During the second testing visit the subject indicated that he was "extremely satisfied" with the comfort of the prosthesis, although at other visits he expressed dissatisfaction due to the weight of the hand. Although we attempted to quantify amount of pro/supination, it was unclear how much was true motion and how much was compensatory. The subject was pleased with the socket, and inquired about using the socket for his body powered prosthesis.

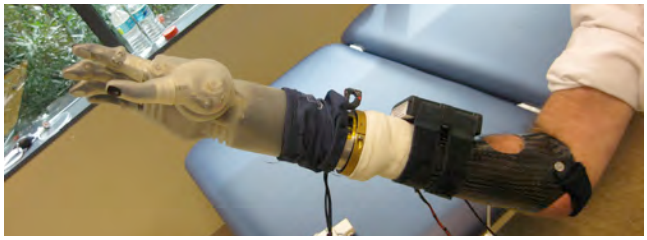

Figure 3: The completed arm with DEKA hand, and user display mounted on the forearm.

DISCUSSION

The dynamic load strap is a promising method to provide a comfortable dynamic suspension which allows for residual pronation and supination. During this case, we used a manual pump to inflate and adjust the dynamic load strap. In the future it would be possible to use an automated control mechanism to inflate after donning, deflate for doffing, and adjust tension depending on loads or movement within the socket.

The subject was very pleased with the overall socket design, including the dynamic load strap. Since this was part of a study of the complete DEKA Arm system, much of the data collected did not isolate the contribution due to the socket design or dynamic load strap.

Future work will evaluate the extent to which a dynamic load strap improves range of motion at the transradial level. There may be other appropriate uses for dynamic load straps in transhumeral or scapulothoracic prostheses.

ACKNOWLEDGEMENTS

This research was supported by VA RR&D, VA RR&D A6780 and VA RR&D A6780I DEKA's support of the VA optimization studies was sponsored by the Defense Advanced Research Projects Agency and the U.S. Army Research Office. The information in this manuscript does not necessarily reflect the position or policy of the government; no official endorsement should be inferred.

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COMPREHENSIVE ARM PROSTHESIS AND REHABILITATION OUTCOMES QUESTIONNAIRE (CAPROQ)

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ABSTRACT

Comprehensive Arm Prosthesis and Rehabilitation Outcomes Questionnaire (CAPROQ) is a questionnaire that is administered to Advanced Arm Dynamic (AAD) patients at various intervals during their prosthetic fitting experience and rehabilitation process. The CAPROQ questionnaire is intended to serve as an opportunity for the patient to provide information and feedback regarding the overall satisfaction they have with their prosthesis and rehabilitation, as well as the function of their prosthesis. The feedback provided is intended to help improve care for AAD patients, assist future prosthetic users and potentially provide feedback to prosthetic manufacturers.

The CAPROQ questionnaire takes anywhere from 30 minutes to one hour to complete and is divided into seven sections which include the following: background and demographics, prosthetic history, primary prosthesis satisfaction and comfort, pain, rehabilitation services, ADL/IADL completion, and satisfaction with AAD staff. A historical overview of CAPROQ will be discussed, as well as how the administrative process works. A general view of the seven sections and results of over 100 questionnaires completed will also be provided.

Thursday, 18 August 2011

Presentation of Posters - 3:00 pm - Foyer of the Wu Conference Centre

Poster Title	Presenter
Design of a Myoelectric Controller for a Multi-DOF Prosthetic Hand Based on Principal Component Analysis	Jacob Segil
Development of a Pattern Recognition-based Myoelectric Transhumeral Prosthesis with Multifunctional Simultaneous Control Using a Model-driven Approach for Mechatronic Systems	Alexander Boschmann
The Prosthetics Needs of Farmers and Ranchers with Upper-Limb Amputations	Craig Heckathorne
A Comparison Between Three Pattern Recognition Algorithms for Decoding Finger Movements Using Surface EMG	Christian Antfolk
Newborn Child with a Rare Disorder Resulting in Transhumeral Amputation, Fitted with a Prosthesis at the Age of 6 Months	Benedikte Holck
The Myoelectric, Fluidic Driven Elbow Orthosis - OrthoJacket	Stefan Schulz
Identification of Patterns in Upper Limb Prosthetic Usage by Analysis of Visual Attention to Areas of Interest	Florin-Alexandru Popa
Multimodal Input Device with SEMG and Contact Force Sensors	Øyvind Stavdahl
Using A Virtual Reality Environment (VRE) to Facilitate Training with an Advanced Upper Limb Prosthesis	Linda Resnik
Growing Up with Unilateral Congenital Below Elbow Deficienty: A Qualitative Study of Individuals Who Currently Wear an Upper Extremity Prosthesis	Vivian Yip
Movement Onset Detection in Various Positions for State-Based Myovcontrol Scheme	Thomas Lorrain
Assessment and Validation of the UNB Test of Prosthetics Function	Eric Karosan

Thursday, 18 August 2011

Presentation of Posters - 3:00 pm - Foyer of the Wu Conference Centre

Poster Title	Presenter
Design of Hand Prostheses Based on Data Captured During Reaching to Grasp Activities of the Human Hand	Doina Bucur

DESIGN OF A MYOELECTRIC CONTROLLER FOR A MULTI-DOF PROSTHETIC HAND BASED ON PRINCIPAL COMPONENT ANALYSIS

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INTRODUCTION

The goal of this investigation is to develop a multi-degree of freedom (DOF) prosthesis controller that uses myoelectric signals as control inputs and which has been dimensionally optimized using Principal Component Analysis (PCA). Currently available multi-DOF hand prostheses cannot be fully utilized because there are fewer control inputs than the number of degrees of freedom (i.e. – joints) that need to be controlled [1]. Based on work from the field of neuroscience [2] it has been shown that grasping is a 'low dimensional' task. Santello et al. used PCA to quantify the principal components (patterns of joint movements) involved in grasping. It was found that grasping tasks involving a number of everyday items could be described by only two principal components. This implies that multi-DOF hand postures can be controlled using only two degrees of control. Therefore, a PCA-based myoelectric prosthetic hand controller can drive grasping postures with only two independent control sites [3],[4]. This is an encouraging finding since current clinical practice indicates two, or three, independent control sites can be located on the residual limb of a typical person with a transradial amputation.

The following paper discusses the design and development of a PCA-based myoelectric prosthetic hand controller. Also, the results of a validation experiment are shared.

METHODS

Design and Development

The design and development of the controller progressed in several distinct steps. The real-time acquisition processing of electromyographic (EMG) signals for two myoelectric sites was developed using standard of care two-site myoelectric control schemes. The PCA algorithm was derived to calculate 15 joint angles of the hand. Several mappings of the EMG signals to the principal component domain were produced. A virtual hand with 15 degree of freedom and anthropomorphic size (50th percentile male) was designed to be controlled in real time. The following sections discuss design and development of the PCA-based myoelectric hand controller in more detail.

Real-time acquisition and processing of two electromyographic (EMG) signals was developed using standard of care to-site myoelectric control schemes. The raw EMG signal was amplified, band passed, rectified and smoothed using typical 2 site myoelectric technique [1].

Following the EMG acquisition and processing, an inverse Principal Component Analysis (PCA) was performed based on work by Santello et al. Santello et al. had subjects grasp 57 household objects while measuring 15 joint angles in the hand. PCA was performed on the empirical data and produced 14 principal component vectors. (The significance of each principal component is determined by the magnitude of the eigenvalue associated with each principal component vector.) Each principal component vector can be considered a 'pattern of movement' between the 15 joints in the hand. This matrix of principal component vectors is used to calculate the 15 joint angles of the hand as described by Equation 1. Each principal component vector $(\vec{[PC]_{\downarrow n}})$ (is a column vector containing coefficients for each of the 15 joints. The EMG signals are the input to the inverse PCA algorithm and the postural vector made up of 15 joints angles ($\theta_m \theta_m$) is the output. Notice that this algorithm utilizes the dimensionality reduction properties of PCA by only requiring 2 inputs to control a 15 degree of freedom hand. The 15 degree of freedom hand has been effectively reduced to a two DOF system. The inverse PCA based algorithm and its associated dimensionality reduction differentiates this controller from other multi-function myoelectric prosthesis controllers.

$$\left[\vec{PC_1}\ \vec{PC_2} \ldots \vec{PC_{14}}\right] * \begin{bmatrix} EMG_a \\ EMG_B \\ 0 \\ \ldots \\ 0 \end{bmatrix} = \begin{bmatrix} \theta_1 \\ \theta_2 \\ \theta_3 \\ \ldots \\ \theta_{15} \end{bmatrix} \quad (1)$$

As noted above, two EMG signals are used as inputs to the inverse PCA algorithm. However, the EMG signals can be manipulated before the inverse PCA calculation is performed. The EMGs can be mapped on to the principal

component domain by the translation and rotation of the axes corresponding to the myoelectric control signals (Figure 1). The solid red axes are the 1st and 2nd principal components which form the PC-domain. The yellow dashed axes depict a linear orthogonal mapping of the EMG signals to the PC-domain. (The dots are empirical data points from Santello et al.'s grasping trials. The stars are virtual hand postures from this investigation including hand flat (HF), cylindrical prehension (CP), palmar prehension (PP), and lateral prehension (LP).) Notice that the entire PC domain can be described by a linear combination of the EMG signals. This study used both linear orthogonal and non-orthogonal mappings. More complex nonlinear mappings might provide additional benefits and will be studied in the future.

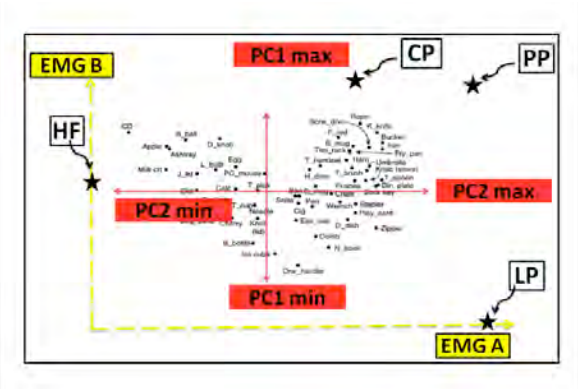

Figure 1: Mapping of EMG signals to PC-domain [2]

The virtual hand design was modeled after the Santello et al. grasping data and built within Labview [5]. The same 15 joints measured by Santello et al. are modeled in the virtual hand (Figure 2). Each digit has two degrees of freedom (the MCP and PIP joints), the thumb has 4 degrees of freedom (the MCP joint, PIP joint, abduction, and rotation), and the middle, ring, and little fingers each have an abduction degree of freedom. The size of the virtual hand and joint axes were modeled after a 50th percentile male hand [6], [7]. A neutral hand posture was derived empirically and is used as the relaxed input to the virtual hand.

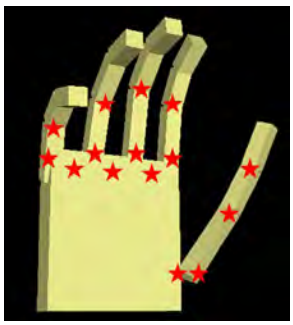

Figure 2: Virtual hand model with 15 articulating joints

Validation Experiment

A validation experiment was performed in order to quantify the effectiveness of the PCA based myoelectric controller. Four postures were commanded and the accuracy of the controlled hand compared to the commanded hand was recorded. Also, different mappings of the EMG signals to the principal component domain were tested in order to determine a mapping that is the most functional. The following sections describe the testing procedure in more detail.

The subject population consisted of 5 subjects. The subjects were normal intact individuals who were able to understand and follow directions in English assessed by their ability to respond during the recruitment and consent process. Exclusion criteria included any subjects with trauma to the upper-limbs including amputation and/or are not able to understand the procedures. Informed consent was obtained from all subjects. Standard of care myoelectric control equipment including surface electromyography (EMG) sensors was used to obtain myoelectric signals from the subjects. Standard clinical procedures involving palpation of the subject's arm was used to locate the best positions on the arm for the surface EMG sensors. ProControl2 surface electrodes [8] were placed over the flexor carpi radialis muscle and over the extensor carpi radialis muscles. A forearm sleeve was worn to hold the electrodes in place while water was applied to the control site to aid in the measurement.

The following 4 postures were commanded during the experiment: lateral prehension (LP), palmar prehension (PP), cylindrical prehension (CP), and hand flat (HF). Lateral, palmar, and cylindrical prehension are defined as the most commonly used grasps during activities of daily living while hand flat is typically used as the neutral posture for a prosthetic hand [9]. All 15 joint angles in the hand were used to define the posture.

3 different mappings were tested during the validation experiment. All mappings utilized two site myoelectric control. Map 1 translated the myoelectric signals to the bottom left corner of the principal component domain (Figure 1). Lateral prehension is accomplished moving along the PC1-axis, hand flat is accomplished moving along the PC2-axis, and cylindrical/palmar prehension is accomplished by a strong co-contraction. Map 2 translated and rotated the myoelectric signals to envelop the data points in Santello's principal component domain. Map 2 was tested to investigate whether axes formed by the trending directions of the grasping data from Santello's work had significance. Map 3 used a non-orthogonal axis system. The axis system was defined as having hand flat at the origin, lateral prehension along the first axis, and palmar prehension along the second axis. Cylindrical prehension was accomplished by a slight co-contraction while on the second axis.

The testing sessions began with a thorough description of the testing procedure and the written consent of the subject. A practice session then occurred. During the practice session, the subject was allowed to control the virtual hand using myoelectric control to gain practice and familiarity with the testing environment. The subject did not use any of the 3 mappings described above during this trial session in order to prevent any familiarity with any of the maps tested. The gain and thresholds of the myoelectric signals were adjusted to provide the most comfortable testing session.

The testing session consisted of 60 randomized trials. Each trial was a combination of a mapping and a posture giving 12 combinations total (ex: Map 1 – Lateral Prehension). Each combination was tested 5 times. The subject was allowed to stop at any point due to fatigue and/or discomfort. The subject was asked to match the image of the controlled hand to the commanded hand within 10 seconds (Figure 3). The subject was provided both the raw EMG signals as well as the joint accuracy measure (an array of lights indicating the percentage of joint accuracy). If the subject achieved the commanded posture, the trial was stopped. At the end of 10 seconds, the trial terminates and the joint accuracy maximum was recorded.

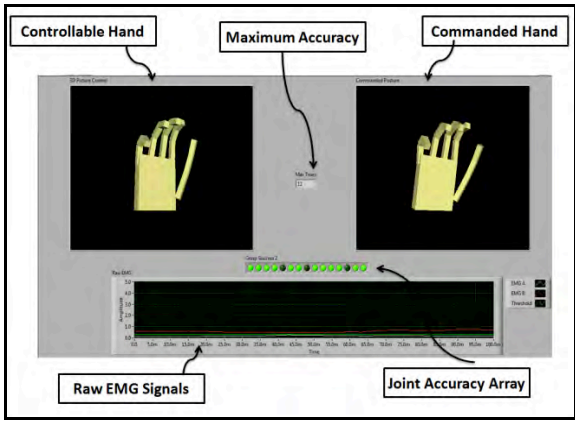

Figure 3: Testing interface

RESULTS

The results of the validation experiment are presented below. A description of the metric studied is followed by the joint accuracy measurements and a description of the statistical methods used. A comparison of the accuracy between postures and between mappings is provided. In conclusion, the favored mapping is found to be more accurate than the other mappings with statistical significance.

Metrics

A maximum joint accuracy percentage was measured during each trial. The joint accuracy metric was defined as the maximum number of joints that are ever simultaneously within the postural envelope. The postural envelope was defined as 25% of the total range of motion of each joint about the target joint angle. This metric is used to compare the accuracy of each posture within each map.

Figure 4 shows the averaged results across all subjects. The accuracy with standard deviation of the four postures is shown for the three maps. It is notable that the accuracy of each posture within any particular mapping is not constant. Also there is a noticeable trend that cylindrical and palmar prehension postures are less accurate than the hand flat and lateral prehension postures across all maps.

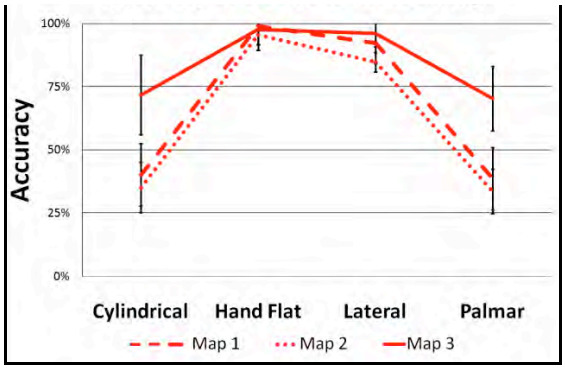

Figure 4: Accuracy of postures across maps

Figure 5 shows the overall accuracy with standard deviation of each mapping across all subjects and postures. Map 3 has a significantly higher accuracy than both Maps 1 and 2.

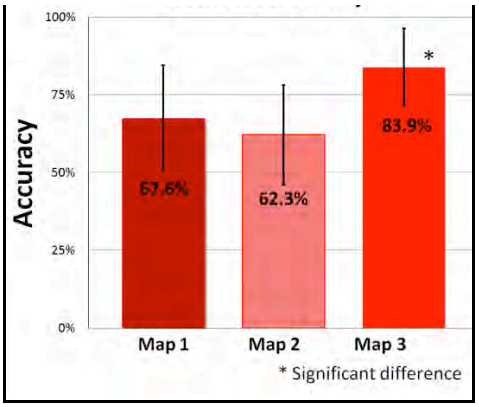

Figure 5: Overall accuracy of maps

In order to verify a significant difference between the accuracy of the three maps an Analysis of Variance (ANOVA) and a Tukey-Kramer test were performed [10]. The null hypothesis for the validation experiment was the following: Map 1, Map 2, and Map 3 produce overall averages that

are equal. The validation experiment (300 sample dataset) produced a p-value (2.6E-6) much less than the acceptable type I error (5E-2) and thereby proving the existence of significant differences between the mappings. The Tukey-Kramer test verified that Map 3 is significantly more accurate than Maps 1 and 2.

Discussion

The following sections discuss the trends found from the results of the validation experiment. The functionality of the maps and the correlation between distance from axes and accuracy is analyzed. Also, the rationale behind testing only maps using two control sites and the clinical implications driving the experimental design are reiterated.

Figure 4 depicts several notable trends. Firstly, it is evident that the accuracy of each posture within a particular mapping is not constant. In other words, some postures are more easily achieved than others for each mapping. Specifically, cylindrical and palmar prehensions are the most difficult postures to achieve. This trend suggests a correlation between the ease of achieving the posture and the distance from a posture and an EMG axis. A co-contraction is necessary to move off of a EMG axis. Especially in Maps 1 and 2, the cylindrical and palmar postures are most distant from the EMG signal axes and are also the least accurate postures. In general, it is found that the accuracy measure is dependent upon the amount of co-contraction necessary for each posture.

Figure 4 also shows where Map 3 proves to be more accurate than Maps 1 and 2. The accuracy of cylindrical and palmar prehension when using Map 3 is over 25% greater than when using Maps 1 and 2. All maps achieve the hand flat and lateral prehension postures easily (with accuracy values above 90%).

Figure 5 depicts an overall accuracy (the accuracy of each map across all postures). This further proves the trend seen in Figure 4 that Map 3 is the most accurate and therefore most functional.

It should be reiterated the rationale behind testing various maps using only two control sites. Standard of care procedures today cite two or three surface myoelectric control sites [11] as the most possible after transradial amputation. This fact constricts the design of a myoelectric controller by preventing the use of multiple (greater than two) control signals. Many technologies have been developed to overcome this constraint including hierarchical control schemes and state-machines. However, the dimensionality reduction provided by PCA allows for continuous morphing between postures as opposed to toggling between distinct states. This characteristic is the most significant advancement made by this project.

FUTURE WORK

The myoelectric controller for a multi-DOF prosthetic hand based on principal component analysis developed in this project will be used in future studies. The validation experiment will be further expanded to include more complicated mappings using both nonlinear and non-orthogonal mappings. Also, more control sites (i.e. 3 or 4) will be implemented and tested using more complicated mappings. It is obvious that more control sites will allow for greater ease of use and functionality. Finally, long term questions that stem from this project focus on dexterous manipulation. More specifically, what are the effects of the higher order principal components and how do they relate to dexterous manipulation? The design of a myoelectric controller for a multi-DOF prosthetic hand based on principal component analysis will hopefully act as a foundation for future studies in this pursuit.

ACKNOWLEDGEMENTS

The authors wish to thank their support VA Eastern Colorado Healthcare System - Denver VAMC and the University of Colorado at Boulder Mechanical Engineering Department.

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DEVELOPMENT OF A PATTERN RECOGNITION-BASED MYOELECTRIC TRANSHUMERAL PROSTHESIS WITH MULTIFUNCTIONAL SIMULTANEOUS CONTROL USING A MODEL-DRIVEN APPROACH FOR MECHATRONIC SYSTEMS

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INTRODUCTION

Modern components and materials in combination with recent pattern recognition methods for electromyographic (EMG) signals enable creating multi-functional arm prostheses with intelligent and user-friendly control [1]. While the usage of pattern recognition of features extracted from EMG signals has proven highly efficient in transradial prostheses [2,3], most current transhumeral prostheses utilize the amplitude of EMG signals from residual arm muscles to control open and close the hand. Co-contracting the muscles usually performs a switch to a different mode like flexion and extension of the elbow, which is cumbersome and does not allow simultaneous movements.

In this paper we describe the systematic development process of an active myoelectric transhumeral prosthesis that allows opening, closing and rotating of the hand with simultaneous extension and flexion of the elbow joint.

Numerous requirements concerning the motion- and security functions have to be considered during the system design process. Therefore we utilize the methodology of model-driven design of mechatronic systems and adapt it to the development of prosthetic systems. Mechatronic models describe both the physical- and the control-engineering model in one integrated model and enable us to design and optimize various aspects of a natural motion sequence from the early phases of the design up to the prototype phase. The result is a prosthesis prototype with an embedded Freescale -based controller. For movement recognition we rely on Support Vector Machines to classify surface EMG signals taken from residual humeral muscles. To validate our approach, a set of experiments was conducted by a transhumeral amputee.

MODEL-BASED DESIGN APPROACH OF MECHATRONIC SYSTEMS

The usage of an integrated development framework supporting the development process from the model to the prototype is crucial in modern active prosthesis development. Especially in the field of mechatronic application the integration of prototyping hardware into the design process is of great importance [4]. The usage of prototyping hardware simplifies the transition from the model to a prototype. It is common to subdivide the model-based design process into three phases: the model-, test rig-, and prototype phase.

In the model phase all system components can be designed and optimized using a virtual model before building a prototype. Different variants of components and functions can be tested by means of simulations. This phase allows the designers to develop the mechanical components in parallel with the actuators, sensor system and electronic functions. The phase results in models able to run under hard real time condition in the test rig phase.

During the test rig phase the already built system components are analysed to determine if they fulfil the performance specifications. Model parameters of the components are identified on a test rig and the dynamic behaviour can be adjusted in the model if necessary. The entire system model will be stepwise adjusted by validated component parameters.

In the prototyping phase the entire system will be analysed and tested. The main focus in this phase is on the examination of effects, which cannot be easily determined using the virtual model. These effects are for example abrasion or friction. Results of these phases form a knowledge base for further development.

APPLICATION TO PROSTHESIS DESIGN

Adapting the model-driven design paradigm to the requirements of prosthetic systems enables the developers to design and optimize all aspects of a natural motion sequence from the early phases of the design up to the prototype phase.

During the development process, mechatronic models are used which combine both the physical- as well as the control engineering models in one integrated model. This model-based approach leads to a considerable reduction of necessary tests. Furthermore, feedback and dynamic system behaviour can be considered in the early design stages.

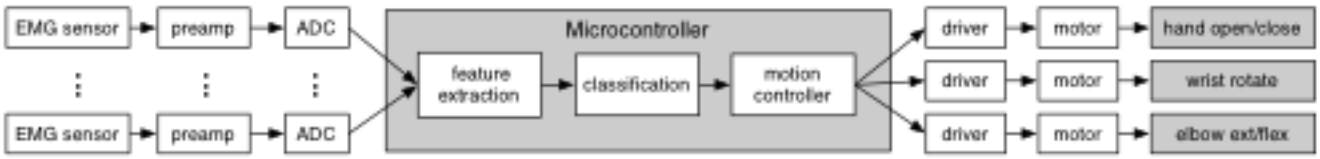


Figure 1: General function. EMG signals are acquired, amplified and digitalized. Feature extraction and classification are implemented on the microcontroller. The motion controller instructs the drivers to perform a movement.

Modelling of prosthesis in CV (Modelling Phase)

Figure 1 shows the function of the prosthesis in principal. It includes all features of a typical mechatronic system consisting of actuators, sensors, a mechanical structure and information processing. All these components have to be developed in an integrative manner.

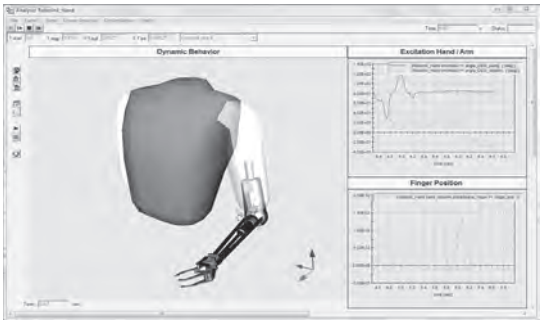


Figure 2: Simulation experiment with 3d animation

The mechanical structure of the prosthesis is modelled as a multi body system, which describes the most important parts of the dynamical behaviour. The information processing unit consists of the feature extraction module, the classifier, and the controller unit for the motion of the prosthesis. Feature extraction and classification are described in the following chapter. Figure 2 shows a simulation experiment of the prosthesis model with time plot and a 3d animation.

Test Rig Phase

The results of the model phase are used as a basis for the construction of the prosthesis. Data of mass, length, forces and torques enable the designer to test the components stepwise on a test rig.

Testing of the controller design that was optimized during the model phase was done with the prototyping system CAMeL-View TestRig [5,6]. With this rapid prototyping system the components of the prosthesis were analysed and set in operation before the prototyping hardware was available. Figure 3 shows a test setup for the controller design. The reference data for the controller can be used from EMG measurement data collected in preceding experiments with test persons.

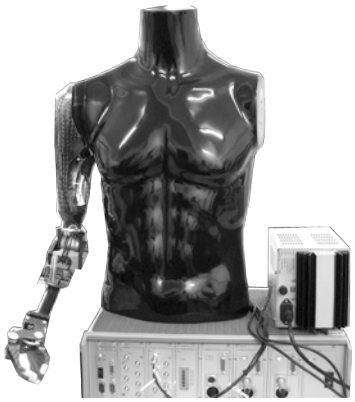
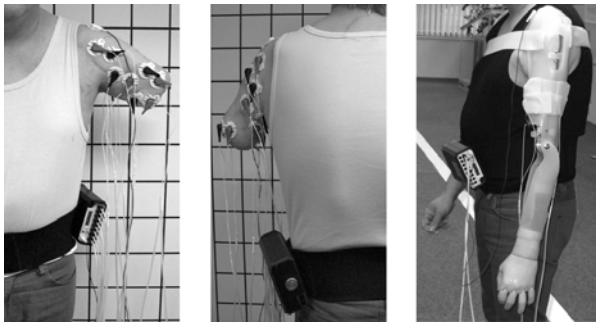


Figure 3: Prosthesis test rig setup

The results of the test rig phase were considered in the model. Identified parameters like bearing friction were compared with model parameters and adjusted accordingly.

Prototype Phase

Fig. 4(c) shows the first prototype of the prosthesis. In the current state of development the system is in an intensive test phase.



(a) (b) (c)

Figure 4: Front (a) and rear (b) view of experimental setup and prosthesis prototype (c)

EMG DATA ACQUISITION, FEATURE EXTRACTION AND CLASSIFICATION

We developed a feature extraction and classification scheme to simultaneously control hand/wrist and elbow movements. It is used in all three phases of the development process.

EMG data acquisition

For EMG data acquisition, we use a Nexus 16 analog digital converter to monitor eight EMG sensor channels with 24-bit resolution at a sampling rate of 1024 Hz. As electrodes we use standard ARBO Ag/AgCl ECG electrodes.

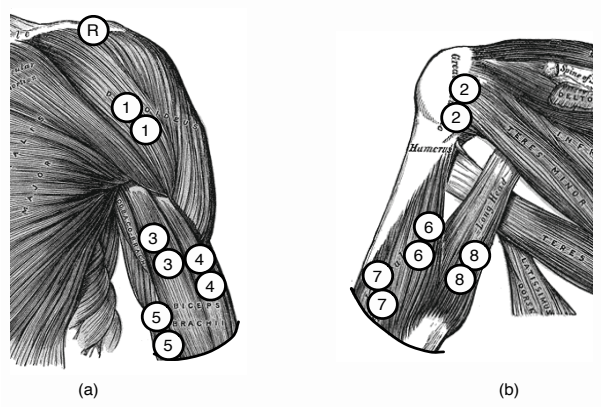


Figure 5: Electrode placing on front (a) and rear (b) arm muscles: 1, 2. M., deltoideus, 3, 4, 5. M. bicept brachii, 6, 7, 8 M. triceps brachii

We have placed the eight electrode pairs on the following arm muscles: M. deltoideus, M. biceps brachii, and M. triceps brachii. Additionally, a reference electrode was placed on the shoulder. The electrode placement scheme is presented in Fig. 5. The exact electrode positions are determined specifically for the test subject to obtain pronounced and reproducible signals.

FEATURE EXTRACTION

Based on the raw EMG signals d_{jkp}, where j denotes the time index, k the channel, and p the movement, we extract features in two steps following the approach presented in [8].

First, the steady state signal starting one second after the beginning of a movement is smoothed by a root mean square (RMS) method with a window size of $w_s = 10$ samples.

The first 100 ms (102 samples at 1024 Hz) of the rectified and smoothed signal are thus given by:

$$d'_{jkp} = [\frac{1}{w_s} \sum_{i=j}^{j+w_s-1} d^2_{ikp}]^{\frac{1}{2}} \quad (1)$$

with j = 1...102. Then, a logarithm-transformed moving average with window size of wf = 20 samples and shift amount of sf = 10 samples is computed from d'jkp. A feature then comprises 10 values and is defined as:

$$f_{l_m kp} = -log(\frac{1}{w_f} \sum_{j=l_m}^{l_m+w_f-1} d'_{jkp}) \quad (2)$$

with lm = 1+(m-1)*sf, and m = 1...10. Two feature vectors are computed: feature vector 1 consisting of features extracted from channels 1 and 2 (20 values), and feature vector 2 consisting of features from channels 3-8 (60 values). This is illustrated in Fig. 6(c).

Movement classification

For EMG signal classification we rely on support vector machines (SVMs) [7]. In our experiments we employ an exhaustive search on SVM's parameters to identify good performing values for C and gamma. An extensive comparison of SVMs to other classifiers for EMG signal classification can be found in [8].

Two classifiers are created during the training phase of the system: SVM 1 from feature vector 1 and SVM 2 from feature vector 2. During the test phase, SVM 1 determines the elbow movement (flexion, extension, relax), while SVM 2 simultaneously decides the hand/wrist movement (hand open/close, pronation, supination, relax). This is illustrated in Fig. 6(d) and (e).

EXPERIMENTAL RESULTS

In this section we report on experiments we have performed to evaluate the system's movement classification performance.

Experiments

In a single experiment run, the test subject had to perform a sequence of six different movements. These movements are hand open and close, pronation and supination of the wrist and extension and flexion of the elbow. In total, 16 experiment runs have been conducted. Each movement starts with a relaxation part of about 4 seconds followed by a contraction part that lasts about 5 seconds, as shown in Fig. 6(a).

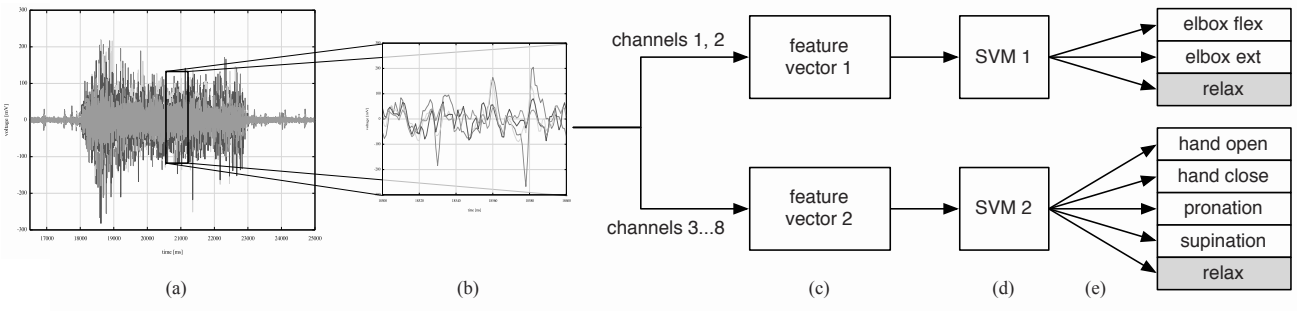


Figure 6: EMG signal processing. Raw signal for all eight channels (a) and 100 ms of the steady state phase (b). Two feature vectors are extracted: one from channels 1 and 2, and one from channels 3-8 (c) and fed into two classifiers (d). Both classifiers determine hand/wrist and elbow movements simultaneously (e).

The EMG signal for the contraction part divides into a one second phase at the onset of the contraction containing the transient components of the EMG signal, and a four seconds steady state phase, which corresponds to a constant force contraction. The steady phase has been used for classification. Features extracted from the 8 odd-numbered trials have been used as training data sets while features from the even-numbered trials were used as training data.

Results

We measure the classification performance of the trained SVM classifier by the classification accuracy, which is defined as:

$$\frac{\text{number of correct classifications}}{\text{total number of classifications}} \times 100\% \quad (3)$$

The classifiers SVM 1 and SVM 2 were used for offline classification of features extracted from the EMG signals. We used 100 ms feature extraction windows with an overlap of 50 ms, resulting in a new prediction every 50 ms. The classification decisions were used to control the virtual prosthesis and the test rig model. Table 1 shows the classification accuracies of the 6 movements. The average accuracy is 90,85%, further investigations will be made to determine whether this accuracy will be sufficient for a satisfying prosthesis operation.

CONCLUSION

In this paper, we have presented an approach to develop an EMG-based transhumeral prosthesis with multifunctional simultaneous control using a three-phased model-driven scheme for mechatronic systems. As a result, a first prototype of the prosthesis was built that allows opening and closing the hand, rotation of the wrist and simultaneous extension and flexion of the elbow joint.

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THE PROSTHETICS NEEDS OF FARMERS AND RANCHERS WITH UPPER-LIMB AMPUTATIONS

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INTRODUCTION

Farming and ranching are among the most hazardous occupations in the United States with many non-fatal accidents resulting in amputation [1], [2]. In addition, those who continue to farm using prostheses are at risk of secondary injuries related to the prosthesis, such as falls, entanglement, and overuse injuries to the intact limb [3]. Furthermore, the hazards of the farm environment are not limited to affecting the adult farmer, but also lead to a higher incidence of amputation among children of farmers than is experienced in children of the general population [4].

Many advances have been made in prosthetics technology since the 1970s, especially with regard to lower-limb prostheses and electric-powered upper-limb prostheses. However, in 2008, the National Institute on Disability and Rehabilitation Research (NIDRR) identified farmers as an underserved population with respect to assistive technology including prosthetics [5]. In response, the Northwestern University Prosthetics-Orthotics Center (NUPOC), as the NIDRR-funded Rehabilitation Engineering Research Center in Prosthetics and Orthotics, partnered with the National AgrAbility Project [6], a program of the U.S. Dept. of Agriculture that provides support services to farmers and ranchers with disabilities, to improve prosthetics options available to farmers and ranchers. The goals of this collaborative project include identification of activities supported by or hindered by use of a prosthesis, provide prosthetics-related educational materials to farmers and ranchers and to the prosthetists who serve them, and to improve prosthetics technology through analysis of failed components and engineering development projects. The project has completed the first phase of a two-part survey of farmers, ranchers, and prosthetists.

METHOD

The first part of the two-part survey was a series of interviews, by phone and in person, to determine specific problems encountered by farmers and ranchers with amputations who were either using prostheses or wanting to use prostheses to enhance their work. Interviews were conducted with 23 individuals with lower-limb amputations, 17 individuals with upper-limb amputations, and 25 prosthetists (across 14 states) who serve farmers and ranchers with amputations. Questions asked of the farmers and ranchers included information about the type and cause of the amputation, type of prosthesis currently being used and history of prosthesis use, types of prosthesis failures or problems experienced, other medical problems and secondary injuries, resources for purchase of prostheses, types of improvements desired, and comments on prosthetics service.

RESULTS

Of the 17 farmers with upper-limb amputations, one had a partial hand amputation, one had a wrist disarticulation, ten had transradial amputations (two bilateral), four had transhumeral amputations (one bilateral), and two had shoulder disarticulations. Thirteen of the farmers had amputations resulting from accidents involving farm equipment.

All of the farmers with amputations distal to the elbow were using a prosthesis at the time of the interview or used a prosthesis for farming before retiring. Only one (transhumeral level) of the six farmers with amputations proximal to the elbow was using a prosthesis at the time of the interview although most had briefly tried using a prosthesis in the early years after their amputation. All of the farmers who use or used a prosthesis in their farm work use cable-actuated body-powered devices. Seven of the farmers had experience with myoelectrically-controlled electric-powered devices but did not utilize them in their farming activities.

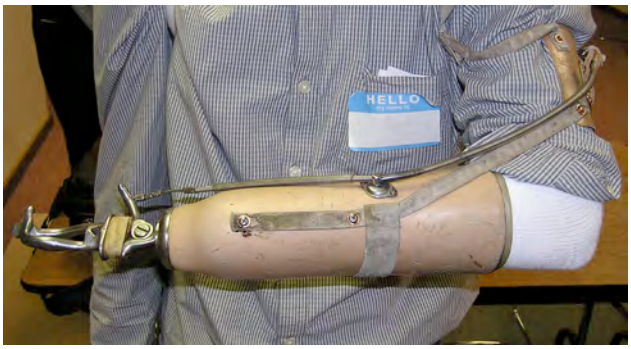

Figure 1: Farmer's body-powered transradial prosthesis

The typical prosthesis for a farmer with a transradial amputation (see Figure 1) incorporated a Dorrance #7 Work Hook, a friction or quick-disconnect wrist, a laminated forearm with laminated or pulled-plastic socket, fabric or rigid hinges, an upper-arm cuff, a figure-of-eight harness of Dacron webbing, and heavy-duty steel cable. One farmer used a TRS GRIP device and one farmer used a polyethylene cable.

DISCUSSION

Farming remains hard work even in the age of mechanized farming and push-button combines. Several farmers described routinely picking up 50 and 100 pound (23 and 45 kg) sacks, climbing silos, handling livestock, connecting and disconnecting farm implements, and numerous maintenance chores.

The number one problem identified by both farmers and prosthetists was durability. Even though prosthetists considered the parts and construction used to be the most appropriate for heavy-duty use, not one farmer thought the devices were durable enough. Common problems mentioned included rapid deterioration of rubber bands due to sunlight, heat and chemicals, failure of wrist units, loosening of the hook from the wrist, breaks in the control cable or pulling of the cable from the fittings, and cracks in the lamination. Most farmers did their own mechanical repairs, and many did not have a back-up prosthesis because of insurance and cost constraints. Concern about durability was the most common reason cited for not using an electric-powered device for farm work, and it is difficult to imagine what kind of repairs a farmer might attempt if a contemporary myoelectric system stopped working.

In addition to the wear and tear of farm work, the farm environment places extraordinary demands on prosthesis performance and construction, including exposure to:

- a wide temperature range
- corrosive or damaging liquids
- airborne particulates
- biological and chemical contaminants

Several farmers mentioned washing the entire prosthesis with soap and hot water to remove dirt and contaminants, a process that would clearly be detrimental to an electric-powered system.

CONCLUSIONS

The interviews are being used to develop a paper and online survey to be administered to a broader representation of farmers and ranchers and prosthetists who serve them. The results of the interviews and broader survey will be used to develop educational materials to support best practices in implementing prostheses for farmers and ranchers and to identify engineering projects to improve component design and construction.

ACKNOWLEDGEMENT

This work was funded by the National Institute on Disability and Rehabilitation Research (NIDRR) of the Department of Education under grant number H133E080009. The opinions contained in this publication are those of the grantee and do not necessarily reflect those of the Department of Education.

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A COMPARISON BETWEEN THREE PATTERN RECOGNITION ALGORITHMS FOR DECODING FINGER MOVEMENTS USING SURFACE EMG

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INTRODUCTION

Classifying surface EMG into different movement types using different pattern recognition algorithms is often used in research in upper limb prosthetics. Several different classifiers have been explored in the literature, however none of them have made it to market for prosthetic hands. Commercially available myoelectric hands are still controlled in a fairly simple open/close manner. As the there are more dexterous hands available for amputees such as the i-Limb by Touchbionics [1], the bebionic by RSLSteeper [2]and soon to be available hands such as the Michelangelo hand by Otto Bock[3], the Vincent hand [4] and the Contineo by Ortocare Innovations [5] the need for robust control algorithms are evident. There has been much research into control of hands using a vast amount of different techniques. Some of the earliest attempts at controlling prosthetic hands using pattern recognition approaches dates back to the 1970's [6].

Using support vector machines and ten commercial Otto Bock electrodes, Bitzer et al. [7] were able to distinguish six classes of movements to control the DLR hand II. Sebelius et al. [8] used a virtual reality hand for training the amputees. Pons et al [9] used virtual hands for training amputees to control the MANUS hand prosthesis using a three bit sequential commands based on EMG. Another method proposed by Nan et al.[10] used five EMG-electrodes and a combination of Bayesian and neural networks to classify both location and motion in a cooking task, classifying six motions and six locations. Xinpu et al. [11] used a new method called SLEX (smooth localised complex exponential) to detect EMG features and a LDA (Linear Discriminant Analysis) to reduce the data set and a MLP (Multi-layer perceptron) network to classify eight wrist motions using six electrodes placed on the forearm of healthy participants. Using four channels of EMG signals Jun-Uk et al. [12] used a wavelet packet transform to extract a feature vector. This vector was subsequently dimensionally reduced using LDA and a multilayer perceptron network was used to classify the outputs to nine hand motions. Ning et al. [13] used a signal processing algorithm for extracting proportional control information for multiple DOF control from EMG signals. A nonnegative matrix factorization (NMF) was used to estimate neural control information from the EMG signals. Cipriani et al. [14]used a four command EMG-classifier and state machines to test different control strategies to command the Cyberhand with 14 able-bodied participants and a knn-classifier to control the Cyberhand in [15]. Tenore et al. [16] decoded individual finger movements (extension/flexion) of each finger (10 movements) using 19 electrodes for a amputee using traditional time-domain features and a multilayer perceptron as a classifier with an accuracy greater than 90%. Shenoy et al. [17] performed an online and an offline study using windowed RMS of the EMG-signal as a feature vector and a Support Vector Machine (SVM) as a classifier to control a 4-DOF robotic arm. Castellini et al [18] used two conditions; still arm (SA) and free arm (FA) to evaluate three different grasps using seven electrodes and ten able-bodied participants using SVMs. User-selected intentional movements were decoded in real time using EMG collected from two sites by Momen et al [18]. Features were extracted using the natural logarithm of RMS values and the feature space was segmented using a fuzzy C-means clustering algorithm. Englehart et al [20] using four channels of EMG compared LDA and MLP approaches using different features in a six class task. Hargrove et al [21] compared classifiers and features using both surface and intramuscular EMG. The preceding work has mostly used surface electromyography. For some indepth reviews on pattern recognition techniques using EMG for control of prosthetic hands see [23],[24].

In this work a comparison between three different pattern recognition algorithms using perhaps the most simple feature set, the Mean Absolute Value, is made.

MATERIAL & METHODS

Ten able-bodied subjects (eight men and two women, aged 25-34) took part in the study. Surface EMG-signals were acquired using an in-house built amplification and data acquisition system. The system acquires 16 channels of EMG, sampled at 2 kHz per channel and with a bandpassfilter between 0.5 Hz and 800 Hz with 16-bit resolution per channel and a gain of 56 dB per channel. A custom-built LabView application (see frontend in Fig. 1) was used to store the data on a PC. A written and visual cue was given as to

which movement the participant was meant to perform. The participants were sitting in front of the computer with their arm resting on a pillow during the time of the experiment.

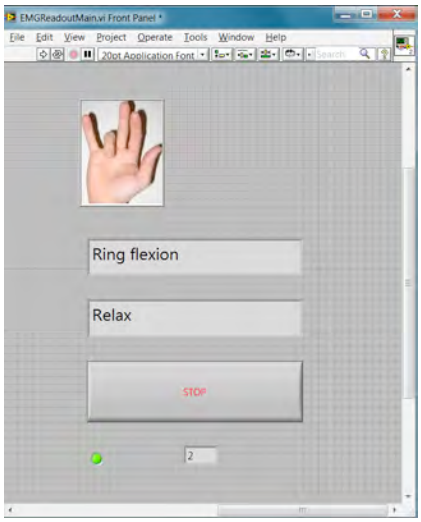

Figure 1: LabView application frontend used when acquiring data

Red Dot Ag/AgCl electrodes (3M Healthcare, Germany) were used in the study. The electrodes were placed on the forearm of the participants as in [8], [15]. Twelve electrodes were placed on the superficial flexor muscles on the volar side of the forearm and four electrodes were placed on the superficial extensor muscles on the dorsal side of the forearm.

The movements used for classification in this study were: thumb flexion, index finger flexion, middle finger flexion, ring finger flexion, little finger flexion, thumb opposition, thumb extension, index finger extension, middle finger extension, ring finger extension, little finger extension, thumb abduction and finally a rest class making up thirteen classes in total. This means flexion and extension of each individual finger as well as thumb adduction/abduction and a rest class. These movements would account for individual control of each digit. In the study after a cue was given the movement was to be held between 4-6 seconds until a rest cue was given. The rest state was of equal length as the movement. Two different datasets each consisting of five repetitions of each movement totalling 60 movements and the rest states were stored on the computer along with the intended class information.

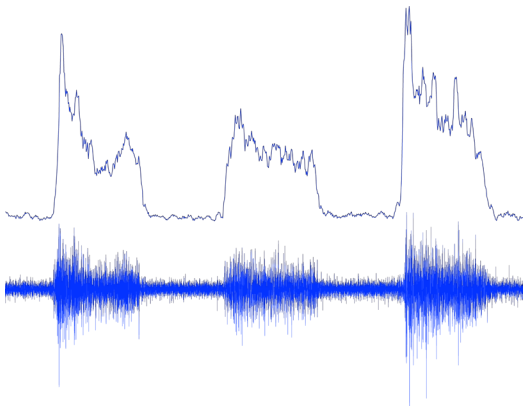

Figure 2: Single channel raw EMG (bottom) and MAV of the same signal (top)

Matlab was used to further process the data. The EMG-data was further bandpassfiltered using a 6^{th} order Butterworth bandpass filter (10-500 Hz) and a 6^{th} order notchfilter (centered at 50 Hz). Each channel of the filtered signals were also normalized. The Mean Absolute Values (MAV) (see top part of Fig. 2) of the filtered EMG-signals were chosen as a feature set. The features were extracted using a window size of 150 ms with a 50 ms overlap. To get an even higher classification accuracy a majority vote filter was used using ten values, five past and five future values. This implies the output of the classifier will be delayed by 250 ms. The delay can be tolerated and the output could still be considered as real-time were it to be applied in such an environment. The whole feature set was chosen as input to the classifiers without cutting rest-periods or performing any additional pre-processing (e.g. PCA).

Three different classifiers were tested: LDA (as has been used in e.g. [20]), k-nn as used by e.g.[8] and a network of multilayer perceptrons as has been used by [16]. All of these classifiers are available in Matlab. The knn classifier used had a k=16 and the Euclidian distance was used as the distance metric. In the MLP network, 16 hidden layer neurons were used and the network was trained using Matlabs scaled conjugate gradient algorithm. Both hidden and output layer neurons had a tansig transfer function. The two datasets were kept separate in the training and testing sessions for all classifiers.

RESULTS

The overall accuracies of the different classifiers can be seen in Table 1. The overall accuracies are not great, but still probably sufficient if using majority voting. Using more features or dimensionality reduction could increase the accuracy of the classifiers.

Table 1: Accuracy of the different classifiers with and without majority vote filtering

Classifier	Accuracy	Majority vote accuracy
LDA	77.66 %	80.66%
knn (k=16)	77.98 %	80.77 %
MLP	79.59 %	82.11 %

DISCUSSION & CONCLUSION

The results show a that there is no great difference between the classifiers, given this problem and feature set. Expanding the feature set would likely yield a higher accuracy, but this would be at the expense of a more complex system. Each of the classifiers would be possible to implement in an embedded system that would be used to control a prosthetic hand. It should also be noted that this setup would lend itself well to be implemented in an embedded system. Calculating the MAV feature is fairly easy computing wise and the filters are not of a high order. Reducing the amount of channels to eight or even lower would also reduce the computing requirements of an embedded system.

Further work would be expanded to also include amputees as they are the ones who would be ultimate user of a classifier such as this in a sophisticated prosthetic hand system.

ACKNOWLEDGEMENTS

This work was sponsored by the Foundation Promobilia, the Swedish Research Council and the Crafoord Foundation.

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NEW BORN CHILD WITH A RARE DISORDER RESULTING IN TRANSHUMERAL AMPUTATION, FITTED WITH A PROSTHESIS AT THE AGE OF 6 MONTHS

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INTRODUCTION

A newborn boy with normal birth weight developed an arterial thrombosis in his left arm (Figure 1) during the labour resulting in a transhumeral amputation (Figure 2).

METHOD AND MATERIALS

The Multidisciplinary Arm Prosthetic Team at the Odense University Hospital, prescribed an arm prosthesis for the child at 6 months of age. The child was fitted with a transhumeral prosthesis, with a soft silicon inner socket (Figure 3). The silicone inner socket was made with shore 65 rolled 1-1,5 mm silicone and baked in an oven for 8 hours. The laminated outer socket had an inbuilt thread, to connect with the special made elbow joint, allowing easy interchanging of the socket.

The special made elbow joint was required to allow ext/flex and wrist joint pro/sup to work in combination with a silicon child's hand. The joint was made from an existing shoulder joint that was modified to allow ext/flex. A tube was connected from the elbow joint to the silicone prefabricated silicone hand, so that the pro/sup movement could be maintained.

Figure 1: Child two hours old

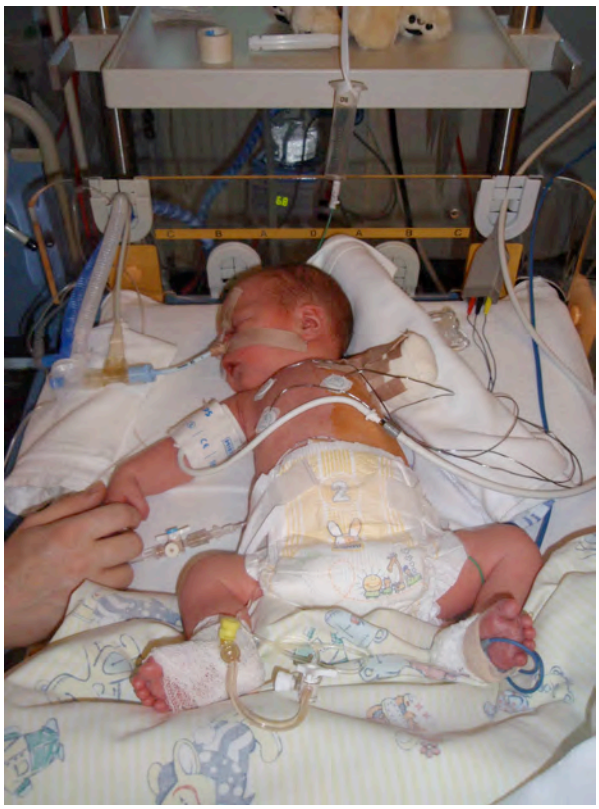

Figure 2: Child three hours old, post op

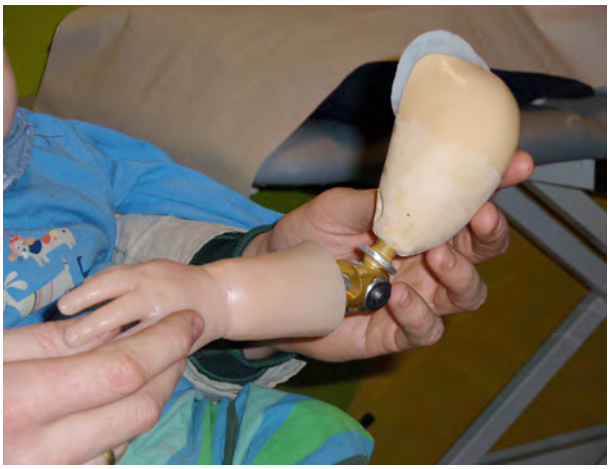

Figure 3: Laminated socket with silicone inner socket, elbow joint and silicone hand

RESULTS

When compared with our earlier experience, the child exhibited a more normal motor neurological development. For example the lying prone with fully extended elbows, allowing for the head to rise (Figure 4). The child also became aware of his prosthesis through normal actions such as biting (the silicon hand for example). Acceptance of the prosthesis was extremely high illustrated by donning the prosthesis early in the morning and removal during the evening. Naptime also included the prosthesis as to allow the prosthesis to become an integrated part of his day. Early on the parents of the child experienced that their child was using the prosthesis without even looking at the hand. Our Team have regular follow-up with all the users of arm prosthesis. The close contact with the family has enabled a good compliance of both parents and child.

Figure 5: Playing

Figure 6 : Happy brothers

Figure 4: Thorax Lift

CONCLUSION

Arterial thrombosis is a rare disorder in newborn children. Our experience is therefore extremely limited in relation to the transhumeral prosthetic outcomes. Despite these limitations, we have found early prosthetic fitting has given the child many advantages. Our team in Odense has over the last 13 years, fitted many trans-radial amputees with a prosthesis from the age of 6 months, so when we were presented to a trans-humeral amputee, our previous experience told us it would be possible.

Children, that have benefited from our early fitting, exhibiting better body symmetry, e.g. to be able to lift thorax, from a prone position, with extended elbows. Biting the hand, shows the children are conscious of the prosthesis and how it feels. Early fitting of prosthesis has allowed improved sit to upraised position. We have also observed that the aforementioned trans-radial amputated children use their prosthesis to rest upon when using their other hand to play.

Though the many follow-up in the multidisciplinary Arm Prosthetic Team at the Odense University Hospital, our experience shows us, that the children become long-time users of the different types of arm prosthesis using from the age of 6 months up to adult life. We see that the children adapt very easy into being two-arms-users. We believe our case study illustrates the advantages of early fitting of arm prosthesis.

ACKNOWLEDGMENT

[1] Anders Peter Højlund, Consultant, Odense University Hospital.

[2] Annette Sørensen, Occupational therapist, Odense University Hospital.

THE HYBRID FLUIDIC DRIVEN UPPER LIMB ORTHOSIS - ORTHOJACKET

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SUMMARY

The project ***OrthoJacket*** (OrthoJacket=orthosis jacket) aims at the development of a modular, active orthosis as a portable system for the upper extremity for high tetraplegic spinal cord injured (SCI). The system combines joint stabilisation, external power from flexible fluidic actuators (FFA) with inherent compliance, a grasping function, realised by functional electrical stimulation (FES) and a natural control system that allows the tetraplegic user to regain independence (see Fig. 1). This article introduces the modular hybrid neuroorthosis OrthoJacket.

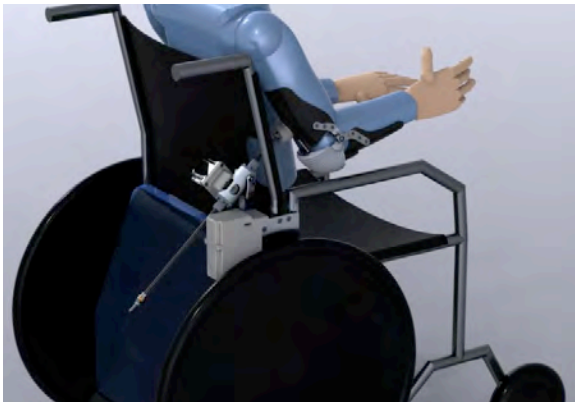

Figure 1: OrthoJacket system, mounted on a wheelchair

INTRODUCTION

Through to the loss of the active movement of the upper extremity, for example a spinal cord injury, patients loose the major part of their autonomy and of their live quality. This leads to a lifelong dependency on caregivers. In the BMBF funded project OrthoJacket a modular, active orthosis for the upper extremity is developed.

The OrthoJacket is primarily intended to be used by high tetraplegic spinal cord injured (SCI) individuals with a cervical lesion (neurological level of lesion C3–C7), which suffer from either complete or incomplete paralysis, eventually with a significant zone of partial preservation or spasticity and spasms. It is aimed as a therapeutic device for enhancement of neuroplasticity in the early rehabilitation phase as well as an assistive device for restoration of persistent functional deficits of the upper extremity. While worn, it will be comfortable and it should be suitable for wearing underneath the clothing.

The primary goal of the wearable orthosis is to improve the paralysed upper extremity function and, thus, to enhance a patient's independence in activities of daily living. The system combines the advantage of orthotics in mechanically stabilising joints together with the possibilities of functional electrical stimulation for activation of paralysed muscles. In patients with limited capacity, for force generation, flexible fluidic actuators are used to support the movement. Thus, the system is not only intended for functional restoration but also for training.

The System consists of an electrically powered shoulder support, a fluid-actuated elbow and a grasping function, realised by functional electrical stimulation (FES). The control of the neuro-orthosis is realised by electromyography (EMG) signals from individually positioned surface electrodes. If there are no measurable EMG-signals, the movement of the orthosis is managed by using a shoulder or neck joystick. OrthoJacket can be used for functional restoration and training at home. By stabilizing the shoulder and the elbow the orthosis relieves the joints, the FES prevents further muscle degeneration and through the active animation joint stiffness is prevented.

GENERAL DESIGN

Depending on the type of the SCI, its location and complexity in the relevant group of patients, the extent of preserved, partially preserved and completely lost functions differs for each patient. Therefore, a modular design is mandatory for the active orthosis to allow for an adaptation and selection of the relevant modules according to the individual status of each user.

For example, a patient with a motor level of C4 needs modules for restoration of the shoulder function, elbow

function and grasping function. In contrast to this, a C5 patient typically requires only a module to restore the hand functions, as shoulder and elbow functions are nearly completely preserved.

A major feature of the basic concept of the new orthosis is the individualisation of the actuator components, where several types of actuators are combined for restoration of the relevant motor functions of the upper extremity. To achieve basic grasping patterns and hand functions FES is applied.

In addition, mechanical and fluidic actuators are used together with FES (see Fig. 2), as it has been shown that FES alone is not sufficient to restore elbow and shoulder functions.

COMPONENTS OF ORTHOJACKET

The concept of OrthoJacket based of three modular parts, which can be used individually or together, depends from patient [1].

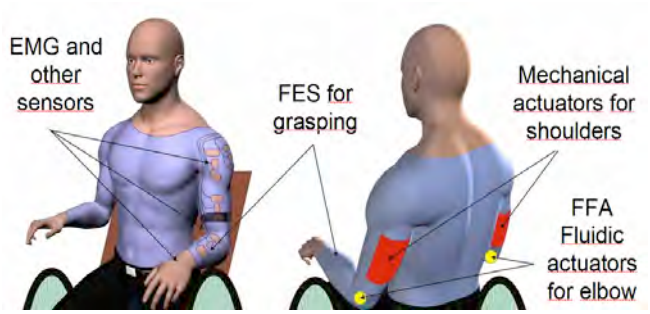

Figure 2: Relevant components of the new orthosis

Hand and wrist module

The movement of the wrist and the grasping function of the hand are achieved by Functional Electrical Stimulation (FES). This type of actuation uses the body muscles to generate the movement. Stimulation is accomplished from outside by special electrodes fixed to the skin above the muscles. Rapid fatigue of the hand is not so critical, because the movements take only very short and no large forces have to be applied [2].

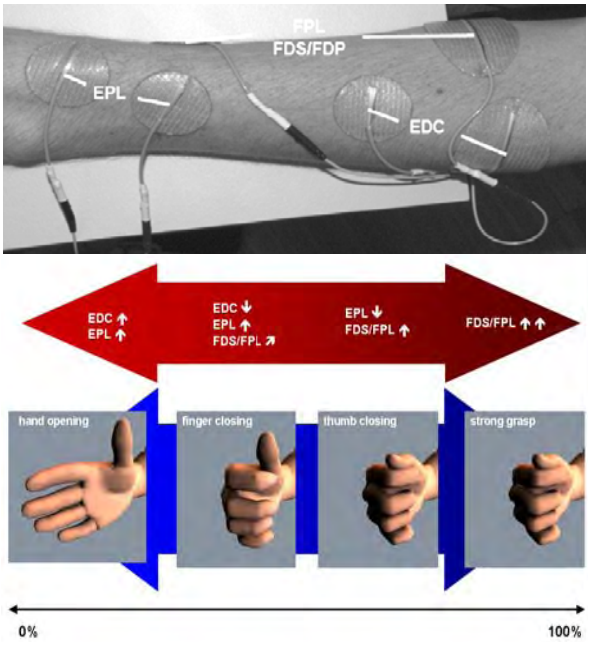

Figure 3: FES electrodes positions and lateral grasp pattern

Grasp function can already be generated by a few surface electrodes, namely three pairs of electrodes for stimulation of the finger extensors (M. ext. digitorum communis EDC), the thumb extensors (M. ext. pollicis longus EPL) and one pair for common stimulation of the finger (M. flex. digitorum superficialis FDS und profundus FDP) and thumb flexors (M. flex. pollicis longus FPL) [3], (see Fig. 3). The muscles controlling the wrist and fingers are located closely to each other in the forearm. Due to the electrode size and inexact positioning of electrodes, not only the relevant muscles, but also adjacent muscles are stimulated. As a result, the wrist direction cannot be adjusted to the desired position. This effect frequently occurs when a simple stimulation system with one electrode pair is applied. The problem is eliminated by the use of several electrode pairs or multi-electrode arrays [4].

Elbow module

At the elbow, the system consists of a lightweight active orthosis that has the potential to be integrated in a jacket. It supports the elbow function up to 100% of the force needed. A series of design studies were needed (see Fig. 4).

Figure 4: Design study of the OrthoJacket powered orthosis with two FFAs on each side of the elbow joint to enable both powered flexion and extension of the elbow (le.) and design with one FFA under use of pressure and vacuum (re.)

For reasons of weight and due to the excellent integrability of FFAs, the orthosis is equipped with these drives that have been developed in the FLUIDHAND project [5][6]. Based on the multibody simulation results, a torque to be applied by the elbow orthosis to move the arm was specified. The actuator meets the required minimum torque amount of 7Nm. The orthosis consists of two composite shells connecting the points of rotation of the actuator with the support area for the upper arm and forearm [7].

Shoulder module

The shoulder function is supported by a linear axle system attached to the wheelchair. The shoulder is actuated by two stepper motors, as the torques to be applied are larger than those at the elbow. The relatively high weight of these drives is compensated by intelligent positioning near the center of rotation of the shoulder system. Shoulder actuation is achieved by a vertical rotation axis for the rotation of the shoulder. Adduction and anteversion are accomplished by an actively driven linear axle fixed to the center of the upper arm to raise the arm (see Fig. 5).

Design of the elbow actuator

As drive a flexible fluid actuator (FFA) is used, because these actuators have a high power density, a small weight, inherent compliance and ensuring safety. Because the actuator is build of several chambers made of film his geometry can be easily adapted to the available space. The new designed fluidic actuator consists of 16 arched and interconnected chambers (see Table 1).

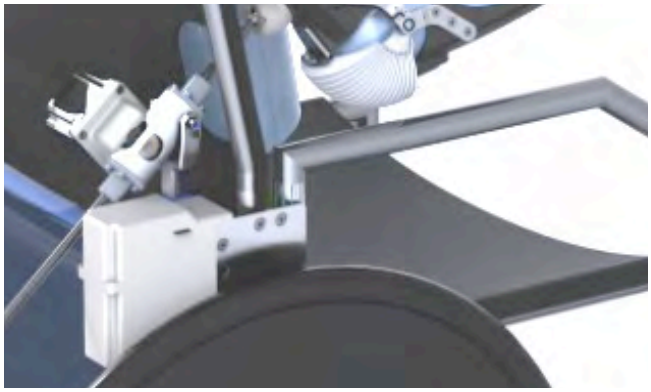

Figure 5: Shoulder module, mounted on a wheelchair

Table 1: datasheet of elbow actuator [7]

Weight	33.2	g
Air volume	16x126 =	mm^3
Air volume	0.020576	l
Thickness at 0 kPa	17	mm
Thickenss at 100 kPa	180 (mechanical stops)	mm
Angle	130 (mechanical stops)	^{0}
Operating pressure	200-300	kPa
Maximum pressure	400	kPa
Burst pressure	960	kPa
Assembly	16 Chambers	
Area per chamber	1737	mm^2

That assumes the shape of a hemisphere under pressure. At each end of a chamber a strap is attached, which serves for the mechanical guide of the actuator. The straps are connected together and rotary associated with the joint axis. Hence, it can be integrated easily in a piece of clothing and hardly interferes with the natural aspect.

For flexion, the actuator is pressurized with an overpressure of up to 400kPa. Extension requires a smaller torque, because it is not necessary to defy gravity. Consequently, 90kPa partial vacuum is sufficient to move the forearm back to the 0_ position. Exact pressure adjustment between -90 and 400kPa is accomplished by a proportional valve. It is located together with the pump and the storage tank for compressed air in a sound-proof container below the seat of the wheelchair [7].

For the measurement of the elbow joint angle a digital angle sensor, based on the Hall Effect, is used. It determines the current angle with a resolution of 12bit.

CONTROL OF THE ORTHOSIS

As the OrthoJacket represents a system with up to 6 degrees of freedom, control is rather complicated for paraplegic patients who have a limited number of usable random signals only. Two different types of random signal sensors are used. If possible, OrthoJacket is controlled by electromyography signals (EMG signals) measured at the skin surface of the patient. There are two approaches to control the orthosis via EMG-signals. If the patient has some remaining voluntary movement in his muscles, for example in the musculus biceps brachii, then the EMG-signal is measured at the muscle the patient wants to move. This kind of control is very intuitive but not always possible, because not every patient hast remaining voluntary movement in the arm or shoulder muscles. For these patients it is still possible to control OrthoJacket via EMG-signals. Here the signals will be taken from muscles with remaining voluntary movement, like the musculus frontalis at the forehead. With a headband with textile EMG-electrodes a frown can be detected and can be translated in a movement of the Orthosis [8].

A second possibility of signal acquisition is to use a joystick fixed on the shoulder or neck. This joystick can detect even smallest movements. As it is impossible to extract a target value for the desired end position from these signals, a speed-proportional control is implemented. When the random signal exceeds a certain patient-specific limit value, the corresponding actuator is activated. The more the current value of the signal exceeds the limit value, the more quickly the orthosis will move. This process is illustrated by an EMG signal in figure 6 below. More details and different modes for the joystick control are described in [9].

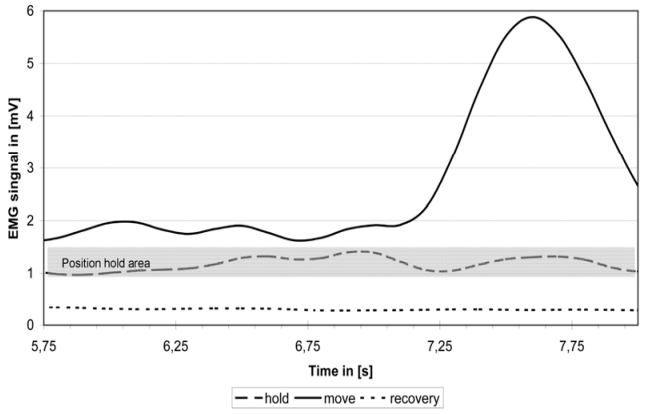

Figure 6: EMG signal at the biceps [9]

EVALUATION

First tests of the system were made with healthy subjects, (Fig. 7). In these tests, it was determined how large the movement space of persons of variable size is and system operation with limbs of variable weight was evaluated. Three persons with complete movability were chosen to represent a very large group of persons. Their weights ranged from 63 to 95kg, their size varied between 1.84 and 1.92m. The results obtained were very good, as you can see on table 2. In case of adduction, the wheelchair to which the system was fixed prevented further rotation to the outside.

Table 2: Results of the evaluation [9]

Patient	Results of the test			
	Weight in [Kg]	*Size in [m]*	*Ante version in [°]*	*Adduction in [°]*
1	63	1.84	0 to 76	-20 to 29
2	84	1.88	0 to 71	-18 to 30
3	95	1.92	0 to 51	-12 to 29

Tests on patients focused on the elbow orthosis. The patient has a lesion below C4 and voluntary movement of the biceps was very difficult. Voluntary activation of the triceps was impossible. With the orthosis flexion and extension of the elbow was between 0° to 90°. The elbow orthosis was controlled by a shoulder joystick. In the patient test, it was checked how intuitive the control of the orthosis is and how reliably it can be moved. When the joystick signal exceeded a certain threshold value, the pressure in the actuator was increased slowly. When the signal dropped below the value, the movement stopped. The results were satisfactory, but also showed that the patient first requires a training phase to learn to control OrthoJacket.

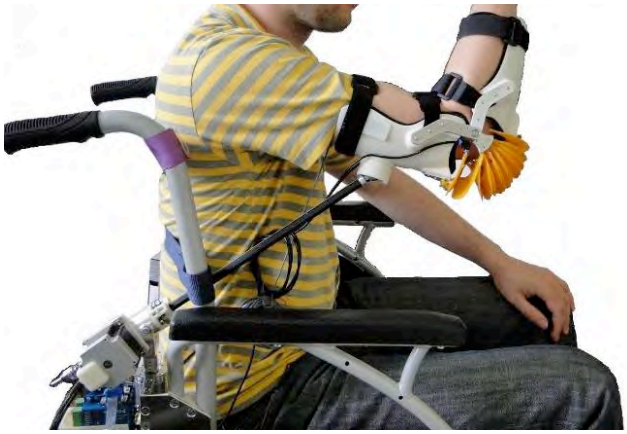

Figure 7: Test of the shoulder- and elbow modules

CONCLUSION

First experiments showed that the elbow orthosis is considered helpful and useful by the patients. Now, the complete OrthoJacket system with the shoulder actuators

remains to be evaluated on healthy subjects and on tetraplegic patients. For this purpose, a test scenario was designed with activities frequently occurring in everyday life. The test person is sitting in his wheelchair in front of a table and wishes to grasp a drinking vessel and move it to his mouth. This movement that is important in everyday life is repeated several times using the different operation modes of the OrthoJacket. With this, it will be tested whether the system will also work reliably in the human environment to get a feedback how the patients has experienced the system.

The design of the OrthoJacket will focus on both functionality and aesthetics. The aesthetics play a major role in the development process, because they affect the patients' acceptance of the new device. Therefore, a distinct focus of the OrthoJacket lies on miniaturisation of the designed components and their integration in textiles.

ACKNOWLEGMENT

This work was supported by the German Federal Ministry of Education and Research (BMBF), funding program "Innovative aids" grant no.01EZ-0774.

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IDENTIFICATION OF PATTERNS IN UPPER LIMB PROSTHETIC USAGE BY ANALYSIS OF VISUAL ATTENTION TO AREAS OF INTEREST

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INTRODUCTION

The performance of upper limb prostheses is based on the ability of users to successfully employ them. This ability is the result of sustained practice and accommodation with the prosthetic device. We are interested in developing a method for assessing the change in visual attention that occurs during the use of upper limb prostheses. The aim of the study is to analyse gaze behaviour for individuals operating upper limb tasks. Previous studies of gaze behaviour suggest that practice in the use of upper limb prostheses determine changes in gaze. These changes can be interpreted as a reduction in cognitive load for the experienced users versus the inexperienced ones [1], i.e. the point of focus moves from the hand to the tool or target object. These results lead to an investigation of the visual attention and its effects on prosthesis usability. We are interested in identifying those factors that are related with changes in gaze behaviour and performance of prosthetic hands. The basic aim is to estimate the cognitive load of users, based on gaze information; see the study by Land et al. [2].

Subjects are asked to perform simulated activities of daily living (ADLs), using the Southampton Hand Assessment Procedure (SHAP). During the tasks, visual attention is monitored and recorded using an eye tracking device placed on the subject's head. Video data of the scene is collected together with information on eye movements, including the coordinates of the point at which the subject is gazing, the point of regard (POR). By using a scheme which codifies where the POR is resting at any one time, the gaze information is then analysed in relation to a series of areas of interest (AOIs). The AOIs are defined by taking into consideration the characteristics of the ADL that is captured in the scene, such as the type of activity and the objects involved. Investigation of the visual attention is then achieved by evaluating the amount of time the POR is fixated in specific AOIs throughout a given ADL. Data processing and analysis are employed using a set of Matlab routines.

BACKGROUND

Studies about gaze behaviour and the influence of visual attention over different upper-limb activities have been conducted in various fields [3,4], but there has been little focus on the evaluation of prosthetic use. A study of driver's gaze behaviour [5] shows that low-frequency changes in the POR, together with longer fixations, are typical for attentive drivers, while high-frequency POR changes and shorter fixation times are typical for drivers distracted by secondary activities. Basically, visual attention patterns change as individuals get more comfortable or skilled in their activities, or when they are less distracted. In these circumstances, lower cognitive load is usually involved. Consequently, it is possible to suggest that there is a correlation between skill level, visual attention and cognitive load. Cognitive load is believed to be an important factor in the learning process of humans. Initially, a task requires a great deal of attention. With time, as the individual becomes more familiar with that particular activity, the task is more efficiently handled. As a result, the individual requires less concentration to achieve the task, hence the lower its cognitive load.

It is known that humans develop models of their body (body schemata), which are used to control their bodies. The body schemata are the result of accumulated proprioceptive feedback from the body. These schemata are affected by an amputation and, subsequently, by the use of a prosthetic device. A study by Mayer et al. [6] shows that changes in body schema are influenced by the time elapsed since the amputation and the start of the prosthesis use. The study concludes that subjects with more experience in using a prosthesis display more overall awareness of their bodies, including of the artificial limb. Body schema acquisition takes time, which means that body awareness is related to the amount of experience the person has in that body configuration. As a consequence, evaluating levels of cognitive load during prosthesis use should reveal this progression.

Our goal is to assess the influence of visual attention over the usability of upper limb prostheses and identify patterns in their usage.

METHODS

Experimental Layout

To this date, baseline experiments have been conducted with anatomically intact individuals performing SHAP tasks. During each trial, subjects wear the head mounted eye tracker (ISCAN). The system records image sequences of the scene and superimposes the POR on each of the frames. Figure 1 illustrates the basic layout of the experiment, as it is captured by the on-head scene camera. During the experiment, the video stream information is captured at a frame rate of 29.97 fps.

Figure 1 - Experimental Layout

Data Processing

The images are processed using the Matlab image processing toolbox and custom software. The goal is to extract the coordinates and boundaries of the AOIs in the scene from the raw videos acquired.

A related study conducted at the University of Salford relies on human interpretation of the AOIs in the scene and is dependent on the level of skill and attention of the observers [7]. Our aim is to develop a software tool capable of identifying the POR and AOI automatically, with minimum human operator input. However, given the complexity of the task, it is not our intention to create an entirely autonomous computer vision system (which would be complex and unnecessary), but a system that uses the operator to give the computer guidance: the initial location of the points of interest and information regarding the layout of the SHAP task, such as where items are likely to be found and what shape or colour they are.

The video analysis involves a long sequence of images. The system needs to identify and track the desired objects within each scene. The task can thus be divided into two parts:

1) Training step (i.e. first frame):
   i. extract and store POR coordinates, based on the crosshair position
   ii. select approximate areas of interest (objects) using mouse input
   iii. extract features for each object within the selected areas of interest (edges, shapes, colour, position)
   iv. based on the pre-compiled feature information, detect all objects of interest in the scene
   v. extract centre of mass (COM) coordinates and a bounding box information for each object
   vi. set up a search window associated with each object, based on its position, shape and size

2) Test steps (i.e. subsequent frames):
   i. extract and store POR coordinates, based on the crosshair position
   ii. using the pre-compiled feature information and associated search windows, detect all objects of interest in the scene
   iii. update the objects' descriptors (colour features, COM, bounding box) based on the current observation
   iv. record speed, acceleration and direction of movement for each COM associated with the tracked objects
   v. set up a new search window associated with each object, based on the shape and size of the object and its COM speed, acceleration and direction of movement

This method basically tracks objects through feature extraction and classification. The classifier aims to identify objects based on a minimal distance between the observed feature vectors and the pre-recorded ones. To date, the package performs the following tasks:

- segmentation of the objects in the scene into AOIs, based on colour and edge information
- identification of the POR within the scene
- evaluation of Euclidian distances between the POR and the COM and bounding boxes of the AOIs in the scene
- object tracking based on colour, shape, COM position and motion information
- evaluation of the time (i.e. number of frames) the POR is fixating at one of the AOIs

The first step towards acquiring useful data for the analysis is segmenting the objects in the scene, which means identifying their boundaries and recognizing them as separate entities. For this purpose, two spatio-temporal methods were employed [8], one based on colour segmentation and the other based on edge segmentation.

The colour based segmentation functions on the basis of similarities in colour intensity. The colour intensity is extracted from the intensity maps of the fundamental colours in the colourmap, the red, green and blue colourmaps [9]. One of the problems with this method is that, under different illumination, a single object can possess different colour attributes. To circumvent this issue, a measure of 'nuance' intensity was implemented. That is, the colour descriptors were derived from the normalized RGB channels, and were called 'redness', 'blueness' and 'greenness'. Two of the newly computed colourmaps are seen at the top of figure 2, together with the original image.

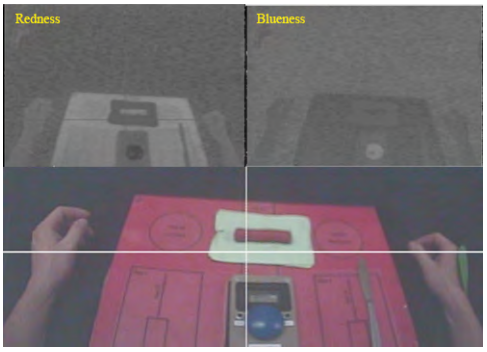

Figure 2 - Red and Blue Nuances Colourmaps

Colour segmentation lacks the ability to discriminate between similarly coloured objects. To avoid this confusion, edge based segmentation is used in conjunction with the colour segmentation. This method is based on detecting sudden changes in intensity, acting as a measure of the second spatial derivative of an image. Following the filtering of the crosshair lines and subsequent application of edge filtering and binarization steps, for different threshold values, an edge image of the scene is acquired.

Figure 3 - Edge Extraction

By restricting the edge detection to both an AOI (either the mouse selection window or the subsequent search window) and a set of colour features (mean intensity and standard deviations for the RGB and normalized RGB channels), the shapes of objects can be accurately approximated, as seen in Figure 4, where the figures on the top are, from left to right: a) convex shape approximation based on the edge detection (see Figure 3 for reference), b) colour mask applied to the convex shape approximation, c) colour blob identification based on window position and colour characteristics (the bounding box of the object is also visible), d) final shape approximation based on the convex image of the identified colour blob. The rectangular box surrounding the area of interest in the original image is the mouse input selection that occurs in the training frame. In the test frames, the box is generated automatically as a search window, based on the COM motion. Figure 4 shows the segmentation and identification process that takes place each frame, for each object of interest (OI). With this approach, combining edge and colour segmentation, successful AOI definition is accomplished.

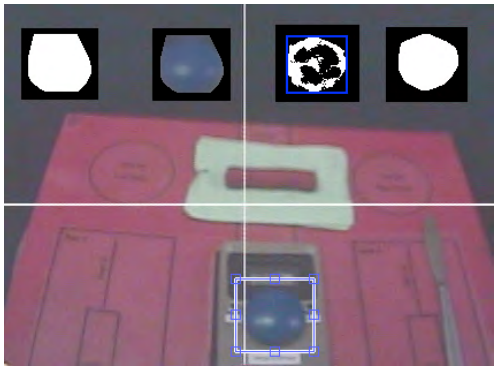

Figure 4 - Assisted Segmentation

Data Analysis

Gaze behaviour is assessed based on the measurement of Euclidian distances between the POR, on one hand, and the COM and bounding boxes of the AOIs, on the other hand. Also, duration of individual fixations is evaluated, by analysing the number of frames the POR is within a certain AOI. These data are sufficient for concluding whether a subject is fixating an object or is just glancing towards it, using its peripheral vision.

RESULTS

Initially, the processed data is represented in a 2D plane. The goal is to assess the relative position of the POR with respect to the COM of each OI in the scene, for each frame of the analysed video. Figure 5 illustrates this representation for a single OI, the blue button from the SHAP kit. For each frame, the COM is plotted at the origin ('stabilized' COM). Consequently, the PORs are represented in rapport with the origin ('adjusted' POR). Their offset to the origin, or relative offset to the COM, is plotted on the vertical and horizontal axes. Each red and blue cross in the figure represent the POR at a specific frame in the analysed video. The blue rectangles surrounding the origin represent the bounding boxes of the OI, i.e. the minimal area rectangles that completely confine

the OI. Separate bounding boxes exist since the object's size and shape is approximated for each separate frame during the processing. The red crosses signal that the POR is inside the bounding box of the OI for a particular frame, suggesting that, for that frame, the subject was fixating the OI.

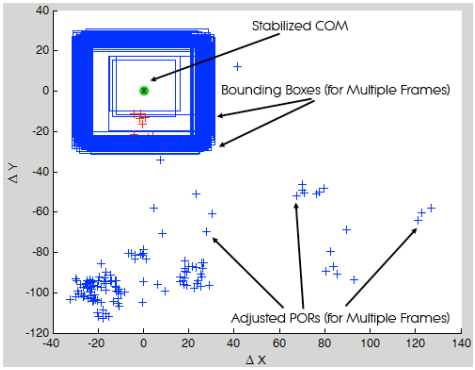

Figure 5 - Relative PORs to COM coordinates

The same data illustrated in Figure 5 is represented in Figure 6 in the form of Euclidian distances between the POR and the COM, for each frame of the analysed scene. The distances are on the vertical axis, while the frame indexes are on the horizontal axis. Red circles mark the frames at which the POR was inside the bounding box of the reference object, again suggesting that the subject was fixating at that respective OI. The illustration is the result of analysing 150 frames or approximately five seconds of recorded video. For comparison, Figure 7 displays the Euclidian distances between the POR and the COM of another tracked object within the scene. In this case, the POR is never within the bounding box of the OI, since none of the distances is represented in red. Note the distance values for this situation, in the rage of 140 to 310 pixels.

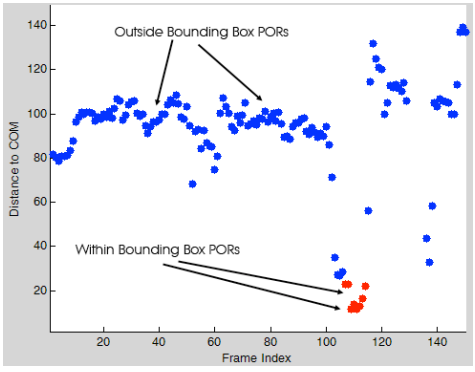

Figure 6 - PORs to COM distances (obj. 1)

DISCUSSION

With this approach, it is possible to evaluate the gaze behaviour by identifying the frames at which the POR is fixated at specific OIs or AOIs. By counting the number of consecutive such frames, the duration of each fixation can be extracted. In Figure 6, the POR was fixated at object 1 (blue button) for a number of 10 consecutive frames. This amounts to 0.66% of the total time analysed (150 frames or ~ 5 seconds). Assumptions can also be made of whether the subject was glancing at a specific object. The concentrated, local minima seen in frames 50 to 60, 115 to 120 and around frame 150, for the second object (see Figure 7), show that the POR suddenly approaches the COM of that OI. This suggests either a glance towards the OI or a focus towards an OI that is in the vicinity (note that around frame 120, POR distances associated with both OIs drop sharply, as caused by the subject fixating object 1).

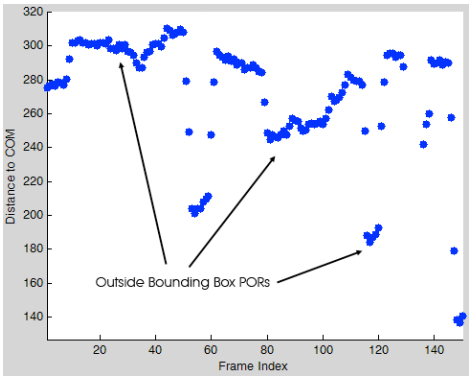

Figure 7 - PORs to COM distances (obj. **2)**

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MULTIMODAL INPUT DEVICE WITH SEMG AND CONTACT FORCE SENSORS

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ABSTRACT

In myoelectric prostheses, movement artifacts are known to impair control performance. This study relates to a novel sensor which measures surface electromyograms (SEMG) as well as contact force at the electrode-skin interface. Its purpose is to explore the in-socket mechanical realities of movement artifacts in order to produce control algorithms that are more robust to said artifacts.

The new sensor includes a commercial SEMG electrode and four surface-mounted force sensors, stacked within a plastic housing. Preliminary experiments in an experienced transradial user showed that sudden lack of control was often caused by electrode lift-off or re-connection. Future work will include algorithms for alleviating these problems.

INTRODUCTION

In myoelectric prostheses, surface electromyogram (SEMG) sensors are located at the very same interface that transfers mechanical load between the prosthesis and the residual limb. Variations in this mechanical load are inevitable during normal use of the device. The accompanying variations in contact force and position between electrodes and residual cause disturbances (artifacts) in the SEMG signals that yield unpredictable electrode output, obscuring the user's motor intent and impairing prosthesis control performance. This is known to be a serious problem to some users of current myoelectric prostheses, to the extent that they choose to turn the prosthesis off in certain situations to avoid unsolicited movements that may cause harm to humans or objects.

The removal of artifacts from EMG signals has been researched extensively. When utilizing SEMG sites on or near the torso, electrocardiogram (ECG) artifacts are of particular interest; see i.e. [1] for a brief review.

When it comes to movement artifacts in prosthesis applications, the literature is considerably scarcer. Lovely et al. pointed out the problem, and suggested an implantable electrode as part of the solution [2]. In prosthesis control systems based on SEMG, movement artifacts are usually attenuated by high-pass filtering the raw SEMG signal with a cut-off frequency of approximately 20 Hz, as suggested in [3]. This filter removes the transient noise induced by most normal upper-limb movements. However, electrode displacement and contact force changes may also induce multiplicative disturbances of relatively low frequencies; this may happen e.g. when a heavy object is being lifted or in certain limb positions, causing the socket, and thus the electrodes, to be pressed harder against the residual, pulled away from it, or simply displaced sideways. Similar effects can be observed if the limb is moved to a new working position, a phenomenon known as the *limb position effect* [4]. This causes the amplitude of the SEMG signal to change, a form of motion artifact that cannot be removed through linear filtering. We propose to include explicit contact force measurements as a supplementary modality in order to identify and attenuate these unwanted phenomena. The resulting device is referred to as a multimodal myoelectric unit (MMU).

MATERIALS AND METHODS

The MMU

Each unit comprises a 13E200 electrode (Otto Bock), which has a built-in preamplifier and produces an output which is roughly proportional to the amplitude of the SEMG. Four FS1500 force sensors (Honeywell), each connected to one of four INA122UA instrumentation amplifiers (Burr Brown Corp.), are employed for contact force measurements. The electrode is coupled to the force sensors with a layer of elastic foam rubber, sandwiched between two semi-rigid plastic sheets, and all parts are eventually stacked within a plastic housing (Figures 1-3; the figures depict an older electrode than the one actually used). The foam rubber acts as a spring that allows the electrode an excursion of up to 3 mm when exposed to contact forces, similar to that of the electrode when mounted the traditional way. Table 1 summarizes the MMU's main characteristics.

The rationale for including four force sensors is as follows. Little is known about the in-socket mechanical realities of movement artifacts. While a single force sensor might enable detection of such states as global electrode lift-off or excessive contact force, with separate sensors in each corner of the device we achieve a joy stick-effect through which we can detect both magnitude and direction of the contact force, and thereby even sideways displacement.

Furthermore, it facilitates the detection of partial lift-off, which may cause the electrode output to saturate and thus hinder all useful control of the prosthesis.

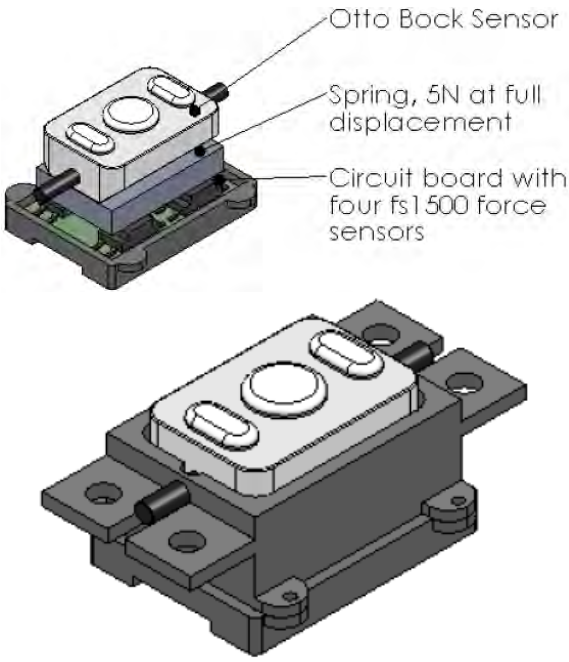

Figure 1: The inner structure of the multimodal device (top); fully assembled device (bottom).

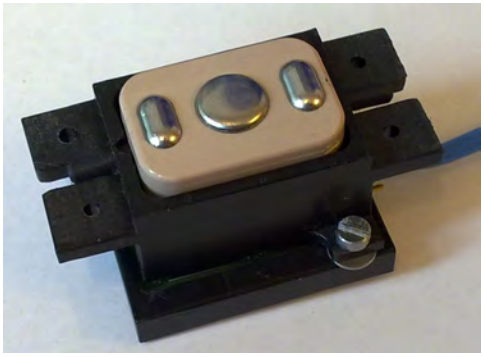

Figure 2: Fully assembled MMU (early version).

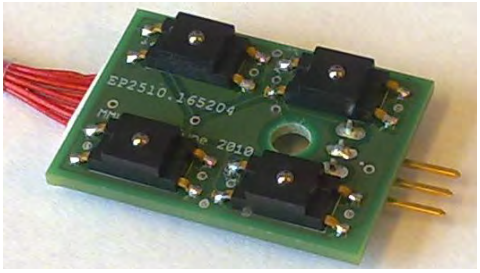

Figure 3: Circuit board with force sensors. The four instrumentation amplifiers are mounted on the opposite side of the board.

Table 1: MMU technical specifications

Parameter	Value
SEMG sensor	13E200 (Otto Bock)
Maximum excursion	3 mm
Contact force at maximum excursion	10 N (approx.)
Force sensors	FS1500 (Honeywell)
Number of force sensors	4
Output signal range (all outputs)	0-5 V
Approximate outer dimensions ex. flanges	W = 25 mm H = 30 mm L = 32 mm

Experimental set-up

Two MMUs were mounted in the socket of a transradial prosthesis with the attachment flanges on the outside of the inner socket. Care was taken to copy the conditions in the user's ordinary prosthesis as closely as possible. All input signals were fed to an analog input/output module, which was connected to a laptop computer via a 5 m USB cable extension. The computer software was configured to sample all MMU signals at 25 Hz and display them on the computer screen in real time. Also, the software is able to produce its own signals and write them back to the prosthesis through analog output channels, emulating electrode output signals. In this preliminary experiment, however, the electrode inputs were simply relayed back to the outputs without modification, in order to have the prosthesis behave in its normal manner.

The computer was set up to log all input and output signals to a storage device, along with video footage recorded during the signal acquisition. The video allowed us to thoroughly study significant events off-line, to establish exactly what happened and what caused it to happen.

The instrumented prosthesis was applied to an experienced transradial user, who was asked to carry out a number of tasks resembling activities of daily living (ADL). The tasks were selected among those reported by the user to frequently cause control problems, and the subject was asked to signal immediately whenever such problems were experienced. The time and type of event was recorded in a written log so that signals and video related to the event could be easily recalled from the database during subsequent investigation.

RESULTS

The user experienced a number of control problems during the experiment, including involuntary opening, failure to open and failure to close.

Figure 4 shows an example of the MMU readout, with the addition of explanatory annotations.

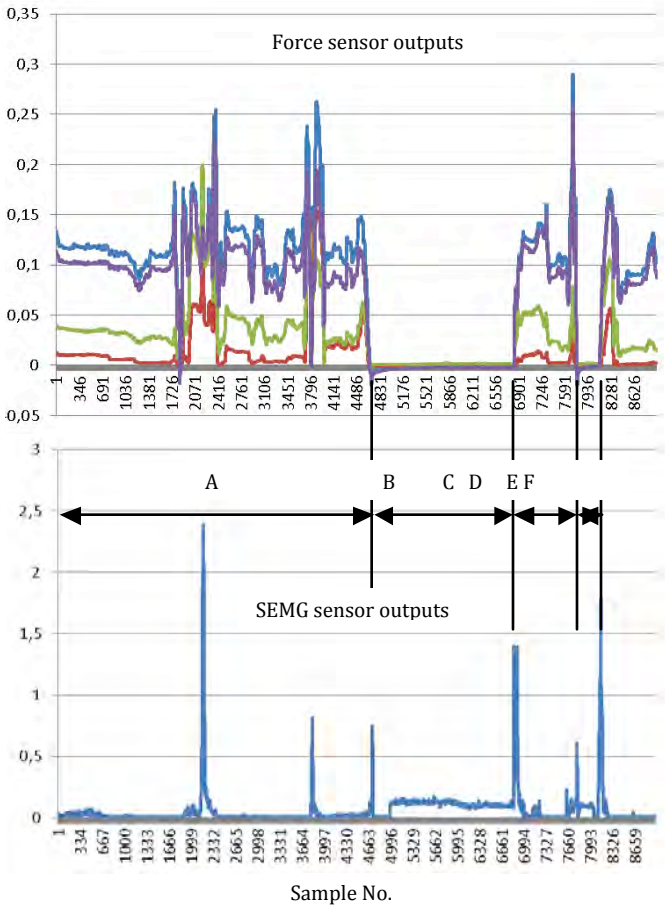

Figure 4: Example of MMU read-out. The graphs show the outputs captured from the extensor MMU, i.e. the one used for opening the terminal device, as the subject was instructed to hold an object with his prosthesis alternately behind his back and in front of him. Significant events have been marked manually off-line.

The events corresponding to the annotations are as follows:

A: Normal operation
B: Arm behind back; failure to open
C: Involuntary opening
D: Normal operation
E: Arm behind back; failure to open
F: Involuntary opening

The following interesting inferences can be made on the basis of these results:

1. The lack of ability to open the terminal device during intervals B and E is caused by total electrode lift-off, as apparent from the corresponding zero valued force signals.
2. The spontaneous opening of the terminal device at C and F are caused by spikes in the electrode output signals. In the force graphs one can see that these spikes occur exactly when the SEMG electrodes re-connect with the residual limb after a period of lift-off.

DISCUSSION

The preliminary results presented in this paper, illustrates that the MMU facilitates detailed studies of various modes of control failure in transradial myoelectric prostheses. This information may be used to optimize socket geometry or mechanical properties of the electrodes, in order to avoid electrode lift-off and similar phenomena that cannot effectively be compensated for through signal processing. Other phenomena, such as changes in signal amplitude due to changed contact force, may in principle be compensated for or reduced through proper processing. The practical applicability of such techniques cannot be established without an extensive amount of data, and ultimately through field testing in the participation domain. This will also require a redesign of the MMU to a smaller form factor and a completely self-contained device.

Although SEMG has been the predominant input signal source for externally powered transradial prostheses, several investigators have demonstrated that even other quantities, used alone or in combination, carry robust information relevant to the user's motor intent. Some relatively recent examples include; myo-pneumatic (pressure) sensors for measuring muscle bulge [5], coupled microphones and accelerometers for acoustic myography and dynamic artifact reduction [6], SEMG combined with near-infrared sensors to quantify local muscle activity through tissue oxygenation [7], and SEMG combined with accelerometers to reduce the position effect [8]. The multimodal device presented here thus fits into a larger family of devices that try to exploit new or supplementary information through *sensor fusion* in order to improve prosthesis control. We note that force (or pressure) sensors have been used by others, but to our best knowledge this is the first attempt at combining high-quality contact force measurements with SEMG.

In compliance with this perspective, the force signals (and any other relevant input information) can be used as full-fledged input signals, not merely for explicit artifact identification and reduction. Such "unified" approaches have been showed to outperform more *ad hoc* methods in certain cases [8] . Whether this approach will yield significant improvement in control performance remains an interesting subject for future research. Obstacles to approaching this goal include the identification of sufficiently general yet realistic

methods for adapting the control parameters to each user, as well as relevant and realistic outcome measures.

ACKNOWLEDGEMENTS

K. Tomm Kristensen of Norsk Teknisk Ortopedi AS, Ottestad, Norway, and Hans Petter Aursand of the Orthopaedic-technical Department, St. Olav's Hospital, University Hospital of Trondheim, Norway, and their respective technical staff, are acknowledged for their invaluable contributions to the identification of relevant subjects, production and adaptation of prosthesis sockets, and for innumerable helpful hints during the course of the work.

This work was funded by the Research Council of Norway under grant 199924/O30.

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USING A VIRTUAL REALITY ENVIRONMENT (VRE) TO FACILITATE TRAINING WITH AN ADVANCED UPPER LIMB PROSTHESIS

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INTRODUCTION

Technological advances in prosthetic design offer dramatically increased possibilities for powered movement. The DEKA Arm system, for example, allows users 10 powered degrees of movement, requiring the user to master multiple controls. No previous prosthetic device has given users control over so many degrees of freedom. Learning to control multiple movements can be a challenge. The majority of controls require the amputee to employ a set of motions, activating muscles that, in most instances, differ from those used to obtain the desired action prior to amputation. When using the foot controls of the DEKA Arm, for example, users must learn to associate motor actions of the feet with specific motor outcomes that are customized for each user.

In the VA Study to Optimize the DEKA Arm we used a prototype Virtual Reality Environment (VRE) program to facilitate motor learning. The VRE program allowed users to practice controlling the arm within a virtual environment, utilizing the same controls used in operation of the actual arm, allowing the user to acclimate to the prosthetic controls prior to using them with the actual prosthesis. This paper describes VRE training used in the VA Study to Optimize the DEKA Arm, and provides qualitative data from a single case study.

METHODS

VRE System

The VRE system used in the VA Study to Optimize the DEKA Arm consisted of a real-time, 3-D avatar that simulated movement of the DEKA Arm system. The avatar consisted of a full torso and head with both upper limbs intact, as well as additional visual information on the selected grip pattern and mode of operation. The VRE provided real time visual feedback on the use of the prosthetic controls, providing the user with information about the aspects and dynamics of movement of the arm for each given command. The avatar in the VRE exhibited the same joint constraints as the prosthetic arm. The avatar could be viewed from the saggittal, coronal and transverse planes, or in a combination of views. The user could zoom in to focus on particular joints, to view the virtual arm from a variety of perspectives. The VRE software used also enables joint motions proximal to the amputation level to be manuevered on the avatar by utilizing slider controls. This feature is applicable to amputation levels including those at the TH level and below and was not used in this case.

Subject

The subject was a 55-year-old male with a left unilateral forequarter amputation secondary to cancer about a year prior to participation in the study. At the time of the study he had been using a body-powered prosthesis approximately 8 hours per day for approximately 3 months. The subject was fit with a DEKA Arm at the shoulder configuration level which was attached via a thermoplastic X-frame socket design with a contra lateral thoracic pad. A pneumatic pressure sensor was attached to the external surface of the contra lateral pad. He was fit with bilateral Inertial Measurement Unit (IMU) foot controls. The subject used foot controls on both feet to provide most control inputs. A pressure sensitive bladder was used to switch between arm and hand modes.

Training Approach

All subjects in our research were trained with the VRE prior to training with the DEKA Arm. All VRE sessions were guided by the study Occupational Therapist. Prior to VRE training, the subject had minimal time utilizing the controls to activate the arm itself. The VRE simulation provided an early experience activating the motor pathways required to operate the controls while observing the motions that they created on the avatar. The VRE training provided an opportunity to introduce the six different grip patterns of the DEKA Arm, and practice opening and closing each grip, memorizing their order of usage. After subjects mastered gross movements they progressed to more complex movement sequences that would be useful to perform basic functional tasks. The subject of this case study had 3 1/2 hours of VRE training over the course of four days, prior to training with the activated DEKA Arm

RESULTS

During his first VRE session, the subject experienced strong sensations of phantom limb movement which subsided

in subsequent sessions. Because of the prototype data transfer approaches used in the version of the DEKA Arm that we were testing, small variations in movement and timing existed between responsiveness of the avatar and the actual DEKA Arm which were noticeable to the subject. Nevertheless, the VRE simulation provided an early experience of activating the motor pathways required to operate the controls and the opportunity to see resultant movements of the virtual arm. By the end of study, the subject was a competent user of the DEKA Arm and performed many functional and recreational activities. The study staff and subject believed that the VRE was a valuable tool in learning to use the DEKA Arm controls.

IMPLICATIONS/CONCLUSION

We concluded that training with the VRE made learning to use a complex set of prosthetic controls easier, however we had no way to test this hypothesis given our study designs. Future studies are needed to evaluate the speed of learning to use complex controls with and without the use of VRE software.

ACKNOWLEDGEMENTS

This research was supported by VA RR&D, VA RR&D A6780 and VA RR&D A6780I. DEKA's support of the VA optimization studies was sponsored by the Defense Advanced Research Projects Agency and the U.S. Army Research Office. The information in this manuscript does not necessar**ily** reflect the position or policy of the government; no official endorsement should be inferred.

The authors acknowledge the valuable work of study coordinator Kate Barnabe, and members of the research team at the VA NYHHS: Nicole Sasson, site Principal Investigator, Ken Breuer, prosthetist, and Maryanne Garbarini, research assistant.

GROWING UP WITH UNILATERAL CONGENITAL BELOW ELBOW DEFICIENCY: A QUALITATIVE STUDY OF INDIVIDUALS WHO CURRENTLY WEAR A PROSTHESIS

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ABSTRACT

With many pieces of literature that debate whether children with upper extremity limb deficiencies should be fitted with upper extremity prostheses (Biddis & Chau, 2007; Biddis & Chau, 2008; James et al., 2006; Wagner, Bagley, James, 2007), it remains uncertain why adults with congenital upper extremity limb loss continue to wear prostheses into adulthood. Our childhood stories contain details of how we have become the persons we are today (Clark, 1993). What childhood experiences have influenced adults with unilateral congenital below elbow deficiency (UCBED) to continue to wear a prosthesis? This study used qualitative methods to capture childhood experiences that have impacted the lives of adults who currently wear a below elbow prosthesis. A phenomenological approach using in-depth narrative interviews of three adults with UCBED targeted 1) positive and negative stories remembered from childhood 2) stories related to use and non-use of the prosthesis, 3) perceived quality of life and identity, and 4) influences to wear a prosthesis. Analyses of these interviews resulted in themes consisting of the participants' backgrounds, growing up and coping with "facts of life", how the individuals continue to cope as adults, the influences to wear a prosthesis, and each individual's personal recommendations for families with a child with UCBED.

MOVEMENT ONSET DETECTION IN VARIOUS POSITIONS FOR STATE-BASED MYO-CONTROL SCHEME

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ABSTRACT

Myoelectric prostheses offer great potential of rehabilitation for amputees. However, a main drawback toward this achievement lies in the control algorithms, which are unable to reliably translate the user's intention to the articulators (prosthesis motors) in an intuitive fashion. This study is part of an on-going project, to develop an alternative approach for myoelectric control, by de-correlating the pattern recognition classes and the actual position of the hand. It focuses on movement onset detection of two degrees of freedom from the hand and wrist complex, in various positions. With 6 channels of surface electromyogram (sEMG) mounted on the forearm, 4 healthy subjects were instructed to perform random sequences of movements with low level of contractions without returning to the rest position. A detection method was developed using a reference buffer and the energy variations in the sEMG. The results show that accurate detection of the movement onset can be achieved regardless of the actual position of the hand within a reasonable delay. This onset detection method is the first step toward a state-based control scheme for myoelectric control. Further work will use this detection method to trigger a classification algorithm and determine the target of the movement.

INTRODUCTION

The surface electromyograms (sEMG) can be non-invasively recorded from the skin surface, and represent the activity of the muscle fibres in the surrounding area. They are easy to acquire, and since they have shown to be an efficient way to control powered prostheses [1], they attracted great interest in the past decades. The state of the art for multi-function prostheses is mainly based on the pattern recognition approach. A large variety of algorithms [2-3] have been investigated for both movement classification and hand/wrist angles estimation, both showing very promising performance [4-5]. However, the pattern recognition approach assumes that different types of motion can be associated to distinguishable and consistent signal patterns in the surface EMG. This assumption, which can be true in very controlled laboratory settings, could be challenged in more realistic conditions. This is one of the main gaps between the academia state-of-the-art and industrial state-of-the-art.

Various commercial powered prostheses are actually available for upper limb amputees; however their control scheme is rather basic due to the lack of reliability of the more advanced methods. They mainly allow controlling two degrees of freedoms, with an unintuitive switching method, for example co-contraction, to switch between the articulated degrees of freedom (DOF).

The aim of this study is to reproduce the performance of such switch based algorithm, without the limitation and restrictions of the co-contractions. A state-based algorithm is being developed providing the control of one degree of freedom at a time, but offering a more natural approach when switching between the different articulated DOFs. This state-based control scheme relies on two main algorithms: the switch detection and the target decision.

The following of this paper introduces the switch detection algorithm and reports its performance on movement onset detection in various positions of the hand and wrist.

METHODS

Subjects

Four healthy subjects (3 males, 1 female) participated in the experiment. All subjects gave their informed consent prior participation to the experiment, and the procedures were approved by the local ethics committee.

Procedures

The experiment focused on 2 degrees of freedom (DoF) of the hand and wrist: Supination/Pronation of the forearm, and Closing/Opening of the hand, as these are the articulated DOFs of the commercial prosthesis the project is working

on. Six pairs of electrodes (Ambu® Neuroline 720 01-K/12, Ambu A/S, Denmark) were mounted around the dominant forearm, equally spaced, one third distal from the elbow joint. The EMG data were recorded in bipolar derivations, amplified with a gain of 2000 (EMG-USB2, OT Bioelectronica, Italy), filtered between 10 and 750Hz, and sampled at 2048 Hz. The reference electrode was placed on the non-dominant forearm. Each experiment consisted of five runs, each performed with a different reference contraction (rest, forearm supinated, forearm pronated, hand closed, hand opened). For each contraction the subject was instructed to maintain the current position, release any previous contraction, and perform the newly instructed contraction at a low force level (~10% MVC). Each run was 2 minutes long and the order of the contractions was randomized. The subject was asked to move at preferred speed, and to keep the contractions for 3 seconds. Five-minute long breaks were observed between runs to minimize fatigue. No feedback was given to the subject to regulate the position or force but visual validation of by experimenter was performed. The wrist and hand kinematics were recorded with a motion capture system (Qualisys Track Manager, Qualisys AB, Gothenburg, Sweden) and 8 ball shaped markers were placed on the subject forearm and hand.

Detection algorithm

The detection algorithm was developed to deal with a continuous stream of data. It is based on a data buffer storing the recent data (i.e. the last 2500ms without positive detection, if available) recorded, and a last event detection time. For each new window (i.e. 40ms), a reliability factor is computed determining how likely the sEMG signal within the current window are in fact a part of a different state than the current one in the reference buffer. This reliability factor is computed according to two main components:

- The time elapsed since the last event detection
- The variance increase in channels between the reference buffer and the current window.

The last event time is the last time a potential switch in the EMG signal has been detected (i.e. reliability factor > 20%). If the previous window had a low reliability factor, the last event time is set as the beginning of the current window.

The time elapsed since the last event was used to avoid false switch detection due do sEMG variability during contractions. A percentage of reliability was computed according to this time value using a time function f_{time} plotted in Fig. 1. This function uses a time base value (i.e. 200ms). This value represents the minimum length to get the maximum reliability factor.

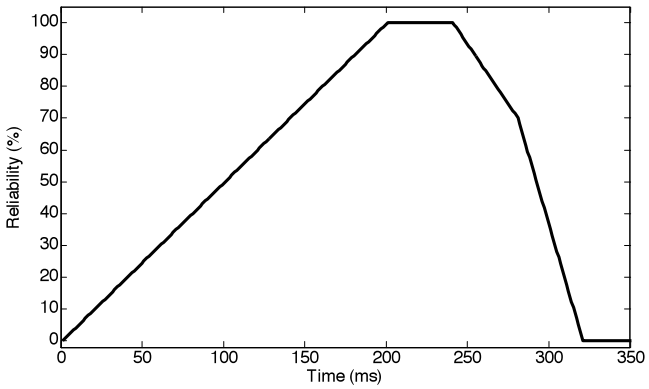

Figure 1: Representation of the time function f_{time} with a time base of 200ms.

To evaluate the variations in the signals compared to the reference state, the data buffer was divided in two parts: the data before the last event time (DBE) and the data after the last event time (DAE). The variation between the DBE and DAE was computed according to the Eq.1. If the left hand side of Eq.1 was larger than 1, the variation coefficient was set to 1 in order to keep its value between 0 and 1:

$$C_{variation} = \frac{1}{N} \sum_{channels} \frac{var(DAE)}{F \times var(DBE)} \quad (1)$$

where N is the number of channels, and F is the coefficient defining the minimum ratio to obtain the maximum value (i.e. F=3 means that the maximum value is reached as soon as the variance ratio is superior to 3).

The final reliability factor was computed as the product of the time component and the variation component, as in Eq.2

$$C_{reliability} = f_{time}(T) * C_{variation} \quad (2)$$

where T is the time since the last event.

The runs were analysed separately as the recordings are not continuous between runs. For each run, the data were down sampled to 1 kHz, and divided into non-overlapping windows of 40ms to simulate the online stream of data. The detection algorithm was run for each window, and a reliability factor for switching state was obtained for each window. The final decision on the detection of a switching event was made by threshold, requiring a reliability value of 80%.

RESULTS

The results for each run were: a true positive rate, a number a false positive per minute, and a latency value for each positive detection. The validation of the detection was done using the motion tracking data. An expected event was placed after each contraction instruction, at the onset of the recorded motion (~10% variation).

Fig.2 illustrates the results of one run

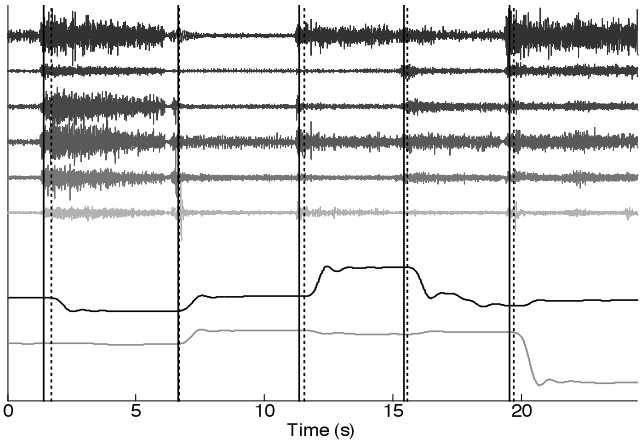

Figure 2: Detection result from one run during 24 seconds. The upper part shows the different sEMG channels. The lower two signals shows the angles from the motion capture system (angle between the fingers and the hand in black, angle between the wrist and the elbow in grey). The solid vertical lines are the detection from the algorithm, the dashed vertical line are the expected events using the motion capture data.

Table 1 summarise the results from all the subjects. It shows that the developed algorithm succeed to detect in average 93% of the performed contractions, maintaining the number of false positive at one per 2 minutes. Although the variability is quite high, the reported latency is in average of -130ms.

Table 1: Detection results by subjects

Subjects	True Positive rate (%)	False Positive (events/min)	Latency (ms)
Subject 1	0.93+/-0.07	0.4+/-0.8	-170+/-190
Subject 2	0.94+/-0.13	0.8+/-1.2	-220+/-200
Subject 3	0.90+/-0.08	0.5+/-0.4	-110+/-310
Subject 4	0.94+/-0.05	0.3+/-0.3	-20+/-410
Average	0.93+/-0.08	0.5+/-0.7	-130+/-290

DISCUSSION

The results of this study shows that the developed algorithm allows to detect more than 90% of the switches performed during the experiments while maintaining a false positive rate of 0.5 per minute.

In addition, the latency reported shows that in average the contraction can be detected prior to the movement. This ensure that the system under development can be responsive enough to be usable in real-time. The variability observed in latency is mainly related to the variation in reference positions. Indeed from a hand in open position with a low level of contraction towards the hand close contraction, the first part of the movement is executed by relaxing the muscles, thus in the first part of the movement there is no increase in the sEMG channels, delaying the detection.

Finally, this detection algorithm is meant to be part of the state-based control scheme. A detection in the sEMG does not imply a movement of the device, as the observed signals will have to correspond to one of the recorded target. Future work will investigate the performance of the entire state-based control scheme, using this detection method as well as the currently under development target detection method. It is expected to provide a more natural and progressive control scheme, yet reliable and clinically applicable.

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ASSESSMENT AND VALIDATION OF THE UNB TEST OF PROSTHETICS FUNCTION

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INTRODUCTION

This paper focuses on the assessment and validation of the UNB test of prosthetics function. There is a constant need to stay up to date with forms of assessment, as assistive devices become more intuitive and precise enabling better control and movement. Due to these advancements in therapy, technology, and education, the UNB test of prosthetics function is subject to redesign in order to better evaluate the usage, function, and training of present day users of prosthetic arms. The aim of this study is to re-evaluate the current protocol to determine any modifications that may be necessary to comply with current standards of practice. The revised test will then be validated through the use of various clinics and users.

THE UNB TEST

The UNB test assesses function with upper limb prostheses and was designed to be simple, quick to administer, taking approximately 30-45 minutes to set up, complete and score a subtest. There are three subtests of 10 tasks allowing test retest which removes concerns over learning effects. The test allows the child to perform his/her task at an unhurried pace. The UNB test is well known clinically and makes an assessment using 10 developmentally based, age-appropriate activities for children aged from 2-13 years. It measures the spontaneity and skill of use with either conventional (body-powered, passive) or myoelectric prostheses, by a trained observer. This particular test was created in 1985, and has been employed by clinicians and researchers, and used as a standard of comparison to newly created evaluations of upper limb prosthetic use [1]. Upper limb prosthetic assessments are integral to the rehabilitation process where maximal functional ability and independence at home, in school, and in the community are the primary concerns.

The new analysis of the UNB test will be broken down to reviewing single components including; culture bias, gender bias, types of grasps, task classification and distribution, progression of child development between age groups, and ceiling effects within age groups. These are variables that influence the functional outcomes of a test. One of the main criticisms of outcome measures that classify categories of function, assessed by clinicians' observed reports, is that categories may not have specific relevance to the individuals' lifestyles or daily routines [2]. Some activities of daily living have changed over the last two decades, and so should the methods of assessing these activities be altered as well.

Choosing the most appropriate outcome measure(s), and having a clear understanding of their strengths and limitations, is important in both clinical and research terms[3] The scoring method will most likely remain as evaluations of skill and spontaneity. An outcome of successful prosthetic use is defined as a person who displays excellent proficiency (skill = 4) and willingness (spontaneity = 4) when using their prosthetic limb. A poor outcome of prosthetic use is defined as a person who displays severe difficulty when attempting to perform the task at hand (skill = 0) and/or refusal to engage (spontaneity = 0) the prosthetic limb to complete the required task.

In the period before 6 years of age there is rapid development and practice of many new skills, whereas after this age the focus tends to be on perfection of skills [4]. Thornby et al [5] also found delay in development of bimanual skills in children with below elbow amputation. Once the modified age ranges have been established, there will be the need to identify any discrepancies in transition phases from one age group to the next. A ceiling effect should be included in each subtest to decrease the likelihood of a younger child performing tasks found in an older age group with ease.

After feedback from clinical practitioners, modifications to the activities for each age-specific subtest will be made while maintaining equality in task distribution between the three subtests of the corresponding age groups. Different patterns of prehensile motion will be classified;

1. Passive use of the hand
2. Maintained grasp of an object
3. Maintenance of the grasp while the person is in motion

4. Repetitive grasp and release of an object during activities
5. Performing grasp and release of the object in any position
6. The ability to grasp and release delicate object
7. The grasp and release of heavy objects

The aim is an even distribution of these activities within and between subtests. The tests will be modified, if necessary, to ensure this distribution. Following this it will be checked for validity. Alterations to the age ranges may be considered, along with the possible broadening of the scope of evaluation through to teens and adults.

CONCLUSION

Accurate and appropriate tests are critical in enabling the correct design of prostheses to be matched to the user or for development of new designs. Its use in prosthetic practice and research enables stake holders to understand more fully their choices for training and prescription. Therefore investigating the validity of a modified UNB test of prosthetic functional outcomes is necessary.

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DESIGN OF HAND PROSTHESES BASED ON DATA CAPTURED DURING REACHING TO GRASP ACTIVITIES OF THE HUMAN HAND

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INTRODUCTION

The design of upper limb prostheses must be done on realistic bases taking into account the present technologies and getting of the best efficiency in their use without ignoring the economical factors.

In authors' opinion, the basic idea in the prosthetic field is that the quality must be measured not only through the technical performances of the prosthetic system but mainly through the performances the wearers get in the daily use of the system.

Naturalness in operation beside cosmetics have to be the major factors in choosing of a certain type of prosthesis.

SPECIFIC ASPECTS ON UPPER LIMB PROSTHESES DESIGN

The prehension mechanisms of upper limb prostheses have to emulate the functional and cosmetic characteristics of the human hand.

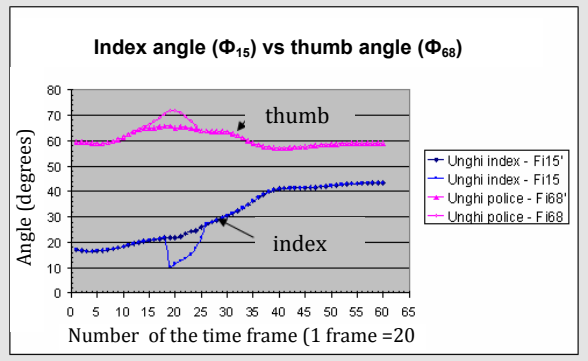

Figure 1

The studies performed in the motion analysis laboratories [1], [2], [3], [4], [9], [11] on the spatial kinematics of the human upper limb, with and without prosthesis, during reaching to grasp activities, pointed out the requirement of designing the prehension mechanisms with the active fingers (index and thumb) having different speeds. This would be in conformity with the biomechanics of the human hand (fig.1), in which the thumb, acting as a stabilizer, has a speed and an angular opening (Φ_{68}) [9] which are less than those of the index (Φ_{15}) which has the main part in the opening/closing activities of the hand (fig.1) [3]. Another finding of these studies has been that regarding the design of the fingers of prostheses which has not to be of anthropomorphic type as the human hand is [3].

Building of the fingers as a link chain, although useful in griping of the objects of complex shapes, asks for sophisticated control systems which give the prosthesis a higher cost these resulting in the danger for the prosthesis of being rejected by the wearer because of the difficulties he encounters in using of it.

DESCRIPTION OF THE EXEPRIMENTAL MODELS OF THE UPPER LIMB PROSTHESES

The four-bar mechanism is often used in the construction of prehension mechanisms, including those of prostheses, as it can be designed to reproduce a wide variety of kinematic conditions [12], [13], [14], [15].

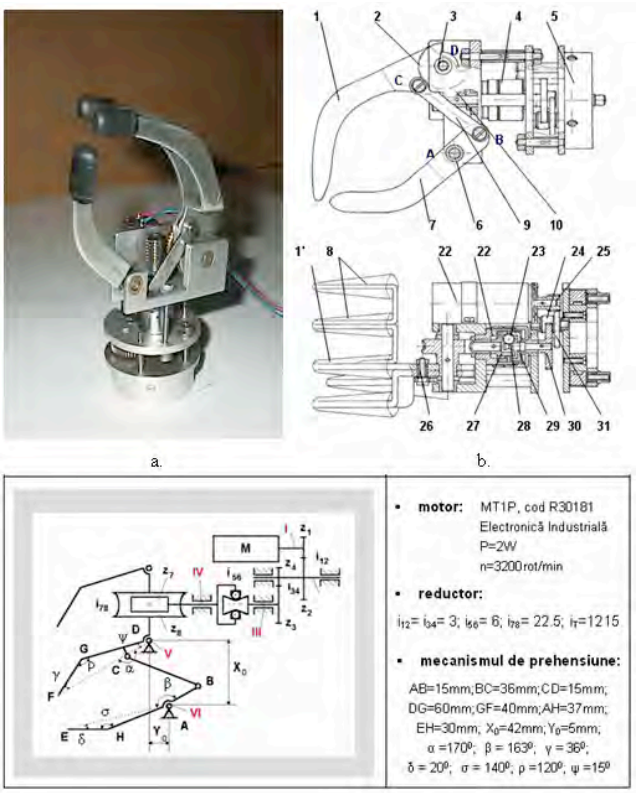

Figure 2

Figure 2 and 3 present such a mechanism and the models designed by the authors.

In the prosthesis from fig.2, a, b, c the fingers 1 and 7 are actuated by building them together with the four-bar mechanism ABCD.

The dimensions of the four-bar mechanism, for the initial adopted solution, in which the fingers move with same speed, are those in fig.2c.

The dc motor 22, is of MT1P type (Electronica Industriala), 2W and 3200 rpm. It actuates the fingers through gears 24,25,30 3, the planetary friction gear 4 and the worm gear 2 which prevent the hand to be opened accidentally.

The opening time is about 2 s and the objects to be manipulated with the prosthesis can be of up to 75 mm diameter.

The construction of this prosthesis is simple and as the motor 22 is placed in the palm, the prosthesis can be use for all levels of amputation including the long forearm amputations. The system 5 allows for passive supination. The control of prosthesis is of myoelectric, on-off type.

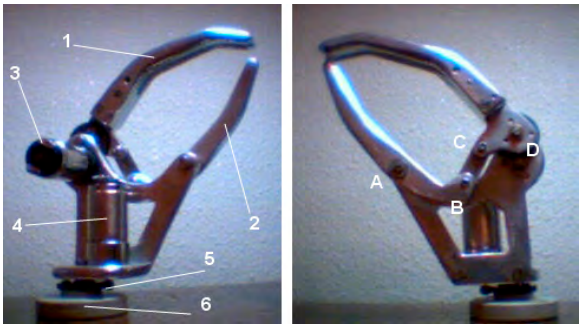

Figure 3

Prosthesis from fig.3 achieves prehension and supination in an active way, using for this two dc motors 3 and 4 (Escape 23L11213P-22) with planetary gears (reduction rate of 128:1), 12V and 7200rpm.

The prosthesis has an opening time of less than 1 s and develops a grip force of about 50-60N. The control of prosthesis is proportional and the electrode are Otto Bock, 13E125=50

PROSTHESES OPTIMISATION AND 3D MODELS

In order the prostheses emulate the kinematics of the human hand the first step was to do the synthesis of the four-bar mechanism [14],[15].

The prosthesis having the rigid fingers shaped corresponding to the resting position of the hand was designed imposing five associated positions: $\varphi_{3i} = \varphi_{3i}(\varphi_{1i})$, i = 1, 5, the values for the angels , Φ_{15} (index) and Φ_{68} (thumb) being chosen from the graph from fig. 1 [1]

The equations projected on the coordinate system of the vectorial equation:

$$\overrightarrow{DC} + \overrightarrow{CB} = \overrightarrow{DA} + \overrightarrow{AB} \quad (1)$$

for the independent contour ABCDA, give a nonlinear system of 10 equations with unknown parameters BC, CD, AD, φ_1, φ_3, φ_{2i}, i = 1, 5:

$$\begin{cases} DC \cdot \cos(\varphi_1 + \varphi_{1i}) + CB \cdot \cos(\varphi_{2i}) = X_A + BA \cdot \cos(\varphi_3 + \varphi_{3i}) \\ DC \cdot \sin(\varphi_1 + \varphi_{1i}) + CB \cdot \sin(\varphi_{2i}) = Y_A + BA \cdot \sin(\varphi_3 + \varphi_{3i}) \end{cases} \quad (2)$$

Solving of the above system of equations was done using a program named *mecanism_proteza* in which were

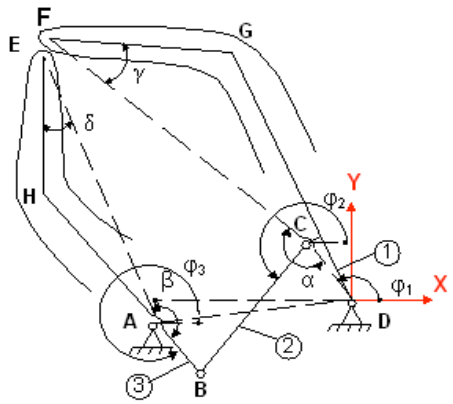

Figure 4

used as initial conditions for BC, CD, AD, φ_1, φ_3, the already known dimensions of the mechanism (fig.2c).

The final dimensions were obtained by multiplying the components of the vector X[i] with adopted valued for AB. The optimal solution for the mechanism was determined with an error ε= 0, 000082, and was as follows:

X[1] = 2.2822812029E+00,
X[2] = 4.6987603654E-01,
X[3] = 2.7611665259E+00
X[4] = -7.980942925E-01X[5] = 1.8380572414E+00

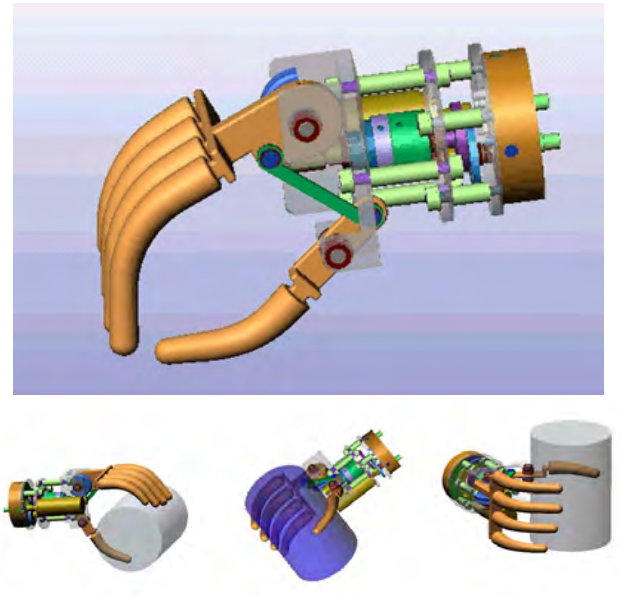

Figure 5

For AB=15 mm it was got:

$$BC = AB \cdot X[1] = 34.23mm;\ CD = AB \cdot X[2] = 7.08mm;$$
$$AD = AB \cdot X[3] = 41.41mm;$$

$$\varphi_3 = X[4] = -0.798rad;\ \varphi_1 = X[5] = 1.838rad$$

The designed mechanism analysed from kinematic point of view allow to obtain for the links CD (index) and AB(thumb) of different angels and speeds, the ratio OM31= ω_3 / ω_1, having the values form the table 1.

TABLE 1
THE RESULTS OF THE KINEMATIC ANALYSIS OF THE REDESIGNED MECHANISM

I	FI11	FI21	FI31	OM1	OM2	OM3	OM31	EPS2	EPS3	EF
	[grd]	[grd]	[grd]	[rad/s]	[rad/s]	rad/s]	-	[rad/s²]	[rad/s²]	[m]
0	105.3129	216.8984	144.1941	-0.262	-0.0355	0.12	-0.4578	-0.0185	-0.0049	0.0543
2	107.3129	217.1603	143.2759	-0.262	-0.0331	0.1206	-0.4603	-0.0189	-0.0046	0.0505
4	109.3129	217.403	142.3531	-0.262	-0.0305	0.1212	-0.4625	-0.0192	-0.0042	0.0468
6	111.3129	217.6263	141.4261	-0.262	-0.028	0.1217	-0.4645	-0.0195	-0.0038	0.043
8	113.3129	217.8297	140.4951	-0.262	-0.0253	0.1222	-0.4664	-0.0197	-0.0034	0.0392
10	115.3129	218.0131	139.5607	-0.262	-0.0227	0.1226	-0.468	-0.02	-0.0029	0.0354
12	117.3129	218.1762	138.6233	-0.262	-0.02	0.123	-0.4694	-0.0202	-0.0024	0.0316
14	119.3129	218.3187	137.6834	-0.262	-0.0173	0.1233	-0.4705	-0.0204	-0.0019	0.0279
16	121.3129	218.4405	136.7415	-0.262	-0.0146	0.1235	-0.4713	-0.0205	-0.0013	0.0242
18	123.3129	218.5414	135.7984	-0.262	-0.0118	0.1236	-0.4718	-0.0207	-0.0007	0.0206
20	125.3129	218.6212	134.8546	-0.262	-0.0091	0.1237	-0.472	-0.0208	0	0.0171
22	127.3129	218.68	133.9108	-0.262	-0.0063	0.1236	-0.4718	-0.0208	0.0007	0.0138
24	129.3129	218.7176	132.9677	-0.262	-0.0035	0.1235	-0.4712	-0.0208	0.0015	0.011
26	131.3129	218.734	132.0262	-0.262	-0.0008	0.1232	-0.4702	-0.0208	0.0024	0.0089
28	133.3129	218.7292	131.0872	-0.262	0.002	0.1228	-0.4688	-0.0208	0.0032	0.0082
30	135.3129	218.7033	130.1513	-0.262	0.0048	0.1223	-0.4669	-0.0206	0.0042	0.0092

CONCLUSIONS

The 3D models of the two prostheses are presented in fig. 5 and 6. The program Solid Works used to build the models incorporates the module Cosmos Works with which can be done kinematic and dynamic studies very useful in the practice of mechanism design.

The theoretical results can be verified on the virtual models which validate the solutions and allows for optimisation of equipment, the overall cost of the final product being minimized because of the low conversion costs being implied.

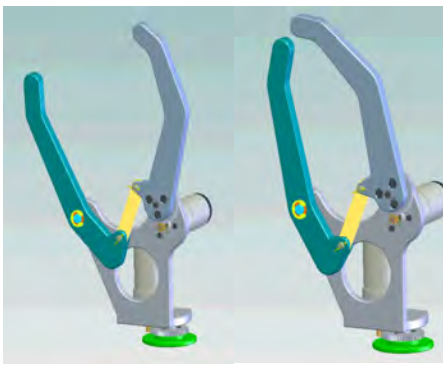

Figure 6

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Friday, 19 August 2011

Wu Conference Centre Auditorium

8:30 AM	**KEYNOTE -Silvestro Micera**

Presentation of Papers

Time	Paper Title	Presenter
9:30 AM	Evolution of the UNB Hand	Yves Losier
9:45 AM	Design Considerations in Upper Extremity Prosthetics	David Altobelli
10:00 AM	***REFRESHMENT BREAK / EXHIBITOR SHOWCASE***	
10:30 AM	Clinical Experiences with the Michelangelo Hand, a Four-Year Review	John M. Miguelez
10:45 AM	The i-LIMB Pulse Hand Compared to the i-LIMB Hand and the DMC plus Hand	Olga van der Niet Otr
11:00 AM	VA Study to Optimize the Gen 2 DEKA Arm: Qualitative Findings	Linda Resnik
11:15 AM	Using the Controller Area Network for Communication Between Prosthesis Sensors and Control Systems	Thomas Idstein
11:30 AM	Bridging the Gap: Ensuring Communication Bus Standard Compatibility with Current Commercially Available Prosthetic Components	Yves Losier
11:45 AM	Towards a Universal Coupler Design for Modern Powered Prostheses	Levi Sutton
12:00 PM	***LUNCH BREAK***	
1:00 PM	Breathable Liner for Transradial Prostheses	Thomas Bertels
1:15 PM	BeBionic Prosthetic Design	Courtney Medynski
1:30 PM	First Experiences with the VINCENT Hand	Stefan Schulz

Friday, 19 August 2011

Presentation of Papers (Continued)

Time	Paper Title	Presenter
1:45 PM	The Electric Terminal Device (ETD) - Case Studies and Evolution	Harold Sears
2:00 PM	Hybridizing body Power & Batteries: Development of the Electromechanical Sure-Lok Cable Control System	Bradley Veatch
2:15 PM	Creative Solutions to Bilateral Upper Extremity Involvement	Debra Latour
2:30 PM	***RAISING THE STANDARD - PANEL DISCUSSION***	
3:00 PM	***CLOSING***	

KEYNOTE:

CONTROLLING HAND PROSTHESES USING PERIPHERAL INTRANEURAL INTERFACES

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INTRODUCTION

The development of more effective approaches to control dexterous hand prostheses is an important area of research that is currently addressed by several research groups. Among the possible solutions to achieve this goal, interfaces with the peripheral nervous system (PNS) and in particular intraneural electrodes can represent an interesting choice. In fact, they can provide an intimate and selective connection with the PNS without increasing in a significant way the invasiveness [1]. In this paper some recent research activities pursued by my team on this topic are briefly summarized.

DECODING OF GRASPING INFORMATION FROM INTRANEURAL SIGNALS

To verify the potentials of intraneural electrodes to decode grasping information, a thin-film longitudinal intrafascicular electrode (tf-LIFE, Fraunhofer Institute for Biomedical Engineering) was implanted in a right-handed male (P.P.) who suffered left arm trans-radial amputation due to a car accident 2 years ago. An algorithm able to sort spikes from the PNS ENG signals was used to verify the possibility to decode grasping information [2].

Results indicate that the combined used of tf-LIFEs and advanced signal processing/stimulation techniques allow identify different grip types usable to control a prosthetic device [2]. The possibility of delivering sensory feedback was also confirmed [3]. Moreover, training and learning capabilities of human-interface interaction, together with a progressive reorganization of the input/output characteristic of the sensorimotor areas previously governing the lost limb were shown.

Finally, the possibility of combining EEG and ENG signals to increase the decoding ability has been also recently shown [4].

DEVELOPMENT OF NOVEL INTERFACES

Current intraneural interfaces can already provide interesting results in terms of decoding and encoding ability but it still necessary to increase their selectivity, stability, and chronic usability. For this reason, we are investigating alternative solutions such as the "self-opening" [5] and "movable" intraneural electrodes [6], which could address some of these issues (see Figure 1).

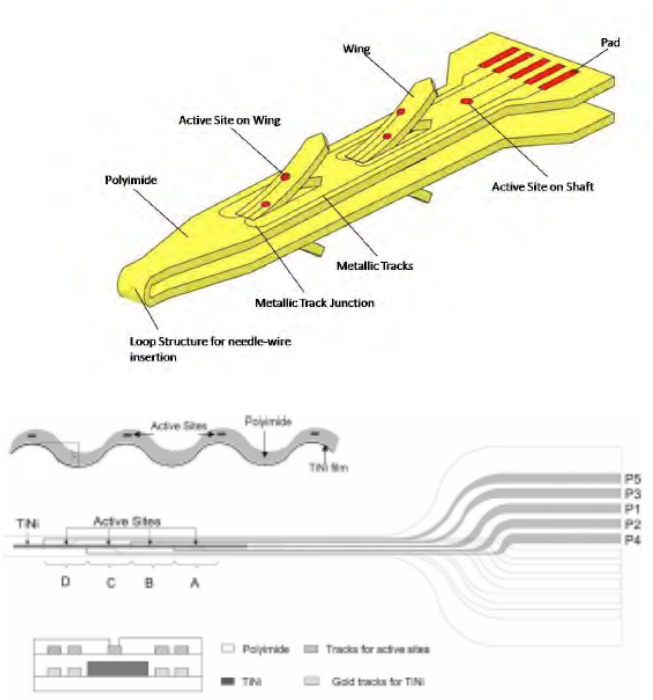

Figure 1: The self-opening (top, [5]) and the movable intrancural electrodes (bottom, [6])

The possibility of developing more effective intraneural interfaces by using hybrid FEM/biophysical models has been also investigated [7].

DISCUSSION AND CONCLUSIONS

Intraneural interfaces with the PNS can represent a suitable way to create a natural and bi-directional link between the nervous system and artificial limbs.

However, additional efforts are necessary to completely characterize the potentials and limits of this approach and its clinical chronic usability. We are currently pursuing several approaches in order to address these issues.

ACKNOWLEDGEMENTS

This work was supported in part by the European Union (EU) within the TIME Project (FP7-ICT – 2007-224012, Transverse, Intrafascicular Multichannel Electrode system for induction of sensation and treatment of phantom limb pain in amputees).

This work was carried out by several students and post-doc who are currently working or worked with me (L. Citi, J. Rigosa, J. Carpaneto, S. Raspopovic, M. Capogrosso, S. Bossi, A. Cutrone, P.N. Sergi). The implantation was carried out by the group of Prof. Paolo Maria Rossini (Campus Biomedico University, Rome, Italy).

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AN OVERVIEW OF THE UNB HAND SYSTEM

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Kevin Englehart[1,2], Peter Kyberd[1,2], and Bernard Hudgins[1,2]

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INTRODUCTION

In 2007, the University of New Brunswick's (UNB) Institute of Biomedical Engineering (IBME) received funding from the Canadian government's Atlantic Innovation Fund program to develop a commercially viable and technologically advanced prosthetic hand system. The 5-year project includes several collaborators namely, the Rehabilitation Institute of Chicago's (RIC) BioMechatronics Development Laboratory, the Université de Moncton's (UdeM) Thin Films and Photonics Research Group, UNB's Applied Nanotechnology Laboratory (ANL), and Liberating Technologies Inc.

The design of this new system, termed the *UNB Hand System*, aims at producing a compact, life-like and affordable hand with a novel cosmetic glove and sensors. The system utilizes a pattern recognition control system based on IBME's previous and ongoing research in the myoelectric control field. This paper provides a general overview of the major components and characteristics of the system.

DESIGN CONSIDERATIONS

Several design factors were outlined at the onset of this project that has helped guide the design and development efforts.

Overall Cost / Affordability

A new prosthetic system will provide benefit to amputees only if it improves the functionality of the prosthetic limb and that it remains affordable. The cost of the system should not be barrier to the acquisition of the system for a user.

Degrees of Freedom (DOF)

A more dextrous hand requires at least two DOF in order to produce various grasps. It was also recognized that increasing the number of DOF within the design would have drawbacks both in terms of weight and fabrication costs. There was therefore a strong motivation to limit how many DOF would be incorporated within the design.

Controllability

Most prosthetic limb users have a limited number of input sources available to control their prosthesis. Although surgical techniques, such as targeted muscle reinnervation [1], can increase the number of available input sources, this procedure will not be available, suitable, or even desirable for many amputees. Pattern recognition control strategies can help maximize the use of available inputs [2] but there are limitations on how much controllability improvement users can gain from such control schemes. In light of this, it was emphasized that the mechanical design of the UNB Hand should only incorporate as many grasps/movements as deemed realistically controllable by most amputees. This criterion also reinforced the argument of limiting the number of DOF used within the mechanical design of the hand.

Modular / Flexible Design

Given that every user has unique requirements (fitting, control, etc), the system must be modular to be easily adaptable to meet the needs of the user. This modular approach is necessary for both the electro-mechanical components and the control options.

SYSTEM OVERVIEW

The system design can be classified as either pertaining to the hand mechanism, cosmetic glove/sensor enhancements, electronic hardware, or control system.

Mechanical Hand Design

The major design considerations included the type and layout of the drivetrain, the number of DOF and the hand's form factor.

Drivetrain Configuration

The hand uses three DC motors, with independent actuation of the index finger, linked actuation of middle, ring and little fingers, and a third motor driving the thumb. The BioMechatronics Development Laboratory at RIC designed a custom gearhead able to provide the necessary force, but basing the design on an off-the-shelf solution to reduce costs.

The flexibility of a prosthetic hand is the result of a trade off between the complexity of a large number of independent motions and simplicity and fewer independent motions. One compromise is to compliantly couple separate degrees of freedom. Within this hand, the compromise is towards reduced costs. So the metacarpophalangeal (MCP) and proximal interphalangeal joints are linked, and all distal interphalangeal joints are fused. There is also no abduction/adduction for the MCP.

The drivetrain for the little, ring and middle finger, used a single actuator where the forces are balanced through two differentials. This provides a conformal grip. The pivots of the differentials were offset, creating an anthropomorphic shape of the hand by positioning the little finger proximal to the middle and ring.

Multi-Degree of Freedom Thumb

For the power and precision based grip forms the thumb is stationary. But unlike current commercial multi-degree of freedom devices the UNB Hand can move the thumb to form a lateral grip without any mechanical assistance from the user. This is accomplished by using a novel mechanism that link this rotation and flexion of the thumb to a single actuator using a cam (Figure 1).

Figure 1: Thumb Mechanism

This cam profile, guides the proximal axis of the thumb, which controls the distal section of the thumb to be positioned in the correct location for each grasp. As the thumb follows the cam it moves from power grip to tripod/precision, to cylindrical/spherical grips, ending opposing the side of the index finger in lateral.

Shape and Component Uniformity

Currently users have to compromise between shape and functionality with passive hands having the most accurate anthropomorphic shape. With the current design of the UNB Hand, the aim was to create a functional 7.5 hand within the envelope of its passive counterpart. This was achieved by 3D scanning a passive glove and creating the design from the outside in (Figure 2).

Due to the limited size of the market, the production of this hand would always be in small batches. It is therefore critical to increase the manufactured quantities within each run, to reduce number of components, and to standardise as many parts as possible. This design achieves this in a number of ways. There is a single size finger (non-handed) with the offset at the MCP, creating the correct anthropomorphic appearance. The entire thumb mechanism, along with all gear trains and drives are design such that a single part can be fitted in both left and right hands.

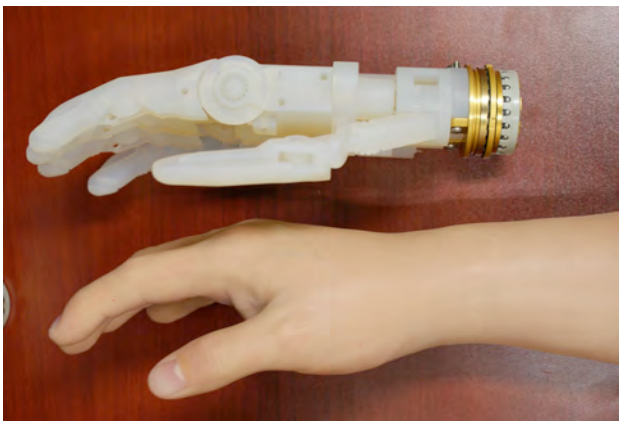

Figure 2: UNB Hand Prototype Comparison with a Passive Cosmetic Glove

<u>Enhancement of Cosmetic Glove Material and Sensors</u>

In addition to addressing the mechanical, electrical, and control aspects of a prosthetic limb system, the project's research and development efforts have included the possibility of improving other features of the design. Specifically, the material properties of prosthetic glove materials and the overall system performance through the inclusion of sensors strategically placed within the mechanical design were highlighted as potential areas of improvement.

Thin Polymer Optical Fibre-based sensors

The Thin Films and Photonics Research Group at UdeM have been investigating thin polymer optical fibres and their potential ability to measure fingertip pressure, lateral slip (slipping through fingers), and distal slip (slipping away from fingers). The polymer optical fibres were shown to be capable of achieving slip detection within a lab setting.

Work has been ongoing to develop and incorporate prototype versions of these sensors within the UNB Hand system.

Prosthetic Glove Material

Improving the properties of prosthetic glove material has also been highlighted as an important aspect of this project. Creating materials capable of exhibiting improved stretch and tear-resistance would not only provide a basis for longer lasting gloves but also improve the performance of the hand and the power efficiency of the system. Towards that goal, UNB's ANL team has focused their efforts on the development of nanocomposite-based materials and subsequent evaluation of the nanofillers' effects on the material properties. Results have shown samples that possess high tear strength and hardness while maintaining high elongation when compared to currently available prosthetic glove materials.

Electronic System Design

The electronic system of the UNB Hand consists of several major components including smart EMG electrode/amplifiers, an advanced myoelectric control unit (AMCU), and a hand controller [3]. All of these components are interconnected via a controller area network (CAN) bus utilizing the prosthetic device communication protocol (PDCP) for information exchange [4]. Each of the major electronic components has unique features while focusing on low power consumption to extend the battery life of the prosthesis.

The smart EMG electrode/amplifier maintains a form factor and power consumption comparable to other commercially available electrodes, as shown in Figure 3, while incorporating a microprocessor and a CAN bus interconnect that can be daisy-chained [5]. The smart electrode also incorporates electrode impedance monitoring capabilities enabling the system to report poor electrode contact or electrode lift-off. This feature may be used to improve overall robustness of the control system.

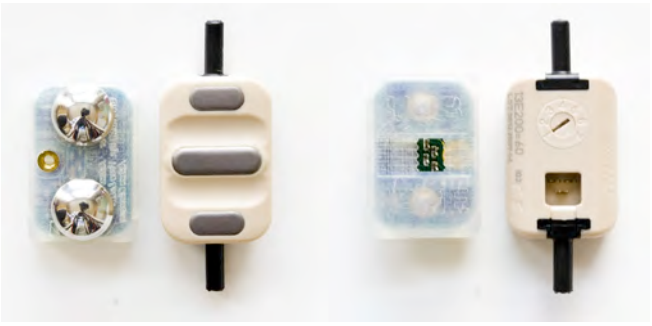

Figure 3: Smart electrode/amplifier and Otto Bock Electrode

The AMCU implements pattern recognition and conventional control systems to generate hand grasp decisions which are passed to the hand controller. The AMCU manages the PDCP communication occurring between the various CAN bus connected nodes on the network. The AMCU is also the communication link between the electronic system in the prosthesis and the host computer responsible for configuring the control system. This communication link to the host computer also enables the EMG signals to be monitored in real-time while the prosthesis is being operated. Finally, the AMCU also incorporates data logging capabilities that will enable logging of system parameters, errors, and failures as well as prosthesis usage information.

The hand controller incorporates the local control system for the hand via the motor controllers and the various sensors in the hand. The hand controller allows the hand to achieve various grasps patterns by monitoring the position of each of the fingers and the thumb and driving each of the motors to the appropriate positions. The hand controller also has the ability to monitor information from force and slip sensors, allowing for grip force control and slip detection and prevention.

Control Paradigm

EMG signal patterns are decoded by the AMCU using well-established pattern recognition techniques developed at UNB. Time-domain features [6] are extracted to maximize information density and classification rate. A linear discriminant classifier is used to perform real-time classification [7]. This clinically validated [8] classification scheme is further enhanced using several pre- and post-processing techniques found to improve control robustness. The final classification results can be used in combination with additional inputs, such as mechanical switches, joysticks, force sensitive resistors and raw EMG signals to support a variety of control schemes. The AMCU also supports common single and dual-site conventional control schemes.

Configuration of AMCU control and engineering parameters is performed through communication with a modified version of UNBs Acquisition and Control Environment (ACE) software package. In addition to hardware configuration, ACE provides data collection and logging capabilities, user training tools, classifier configuration and virtual testing environments.

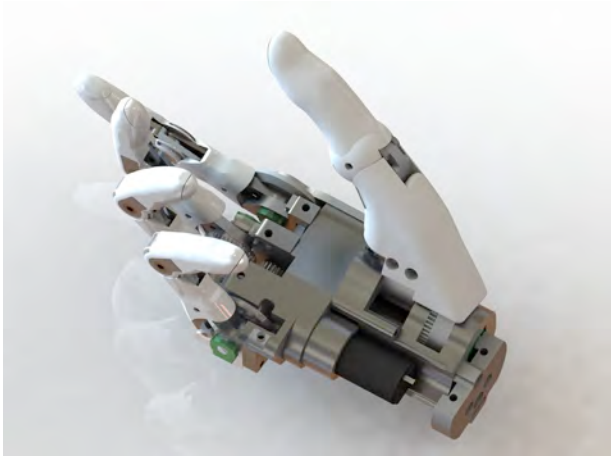

Figure 4: The UNB Hand

CONCLUSION

The UNB Hand System design and development efforts have focused on producing a system that has improved functionality and features over currently available system while striving to remain at an affordable cost. These factors have and will continue to help guide the project through the remaining prototype, testing, and clinical stages.

ACKNOWLEDGMENTS

This project is sponsored by the Atlantic Innovation Fund, the Natural Sciences and Engineering Research Council of Canada, and the New Brunswick Innovation Foundation. Additional funding has been provided by the University of New Brunswick and Liberating Technologies Inc.

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DESIGN CONSIDERATIONS IN UPPER EXTREMITY PROSTHESES

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INTRODUCTION

In 2005 the Defense Advanced Research Projects Agency (DARPA) initiated the "Revolutionizing Prosthetics" program with the goal of dramatically increasing the functionality and capability of upper extremity prosthetic solutions [1]. To support the general goal of restoring near-normal functionality to our wounded servicemembers and other prosthesis users, DARPA particularly focused on increasing the degrees of freedom (DOF) and the capability of the control schemes available to the user. Within the spectrum of research funded as part of the program, our team, led by DEKA Integrated Solutions, was charged with development of a prosthetic arm system that offered dramatic improvements in capability using only non-invasive control schemes.

Mimicking the function of the human arm is a significant engineering challenge. The specifications of the "original equipment" are impressive - 22 degrees of freedom, a vast array of efferent and afferent signals providing actuation, sensation, and feedback/reflexes, combined in a package weighing in at around 7.5 lbs (3.5 kg) and a density of around 1 gm/cm^3 [2,4]. However, advancements in robotic technologies, component miniaturization, manufacturing techniques, microprocessors, sensors, and wireless communications allowed us to develop an advanced upper extremity prosthetic solution.

We employed an iterational, user community-focused design approach for this development effort. Working closely with users, prosthetists, and therapists throughout the process allowed us to capture and quickly implement community feedback. In parallel we focused on solving the difficult engineering problems associated with providing dramatically greater prosthetic arm system capabilities. Where possible, we located our engineering efforts and our clinical studies in the same physical space to facilitate exchanging ideas and rapidly responding to and experiencing the results of our design iterations.

In this paper, we discuss the design approach used for the prosthetic arm system: a general overview of the system characteristics, and a discussion of two specific prosthetic arm system capabilities.

DESIGN CONSIDERATIONS

Our goal was to dramatically expand the capabilities of the prosthesis while enhancing its stability and comfort. Multiple factors needed to be considered in the upper extremity prosthetic design including the arm hardware, control system, power sources, socket interface, and patient control strategies.

The design strategy is schematically represented in Figure 1 and shows the basic elements and pathways for the progression of the design from the first to third generation. Subjective and quantitative data from our engineering team, prosthetists and subjects were analyzed and reviewed before proceeding with the next design iteration.

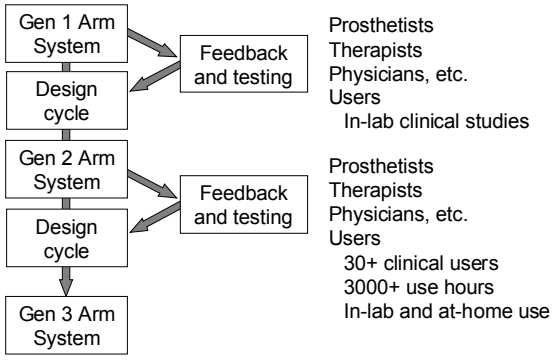


Figure 1. Overall strategy used to advance and prioritize design features through major design iterations.

Our "feedback and testing" process evolved as the design matured. The Gen 1 arm system was used by a smaller set of research subjects over several months. Based on their feedback, substantial improvements were made to the arm system, optimizing the elements of the arm system, the control scheme, and the interface design.

The Gen 2 design was then studied more extensively – increasing the hours of use by research participants, the number of participants, and the environment in which the arm system was used. The Department of Veterans Affairs (VA) established and funded a prosthetic system research team including researchers, prosthetists, and therapists from multiple VA and military centers that joined the clinical study

effort. This team brought feedback from a larger set of users, therapists and prosthetists to the engineering team to support the Gen 3 design effort.

In total, the Gen 2 arm system was used in clinical studies for over 3000 use hours by over 30 users at all configuration levels. Studies were performed at clinical locations at DEKA, Next Step Orthotics and Prosthetics, and the several VA locations. In addition, five study participants were able to take the arm system home for several weeks of use in a non-clinical setting.

The extensive Gen 2 study team provided significant feedback across the entire arm system with insights and suggestions regarding: grip design, joint range of motion; the control system implementation; and the active socket interface. In addition, engineering studies of reliability, capability, and joint use provided additional valuable information incorporated in the Gen 3 design.

GEN 3 SYSTEM OVERVIEW

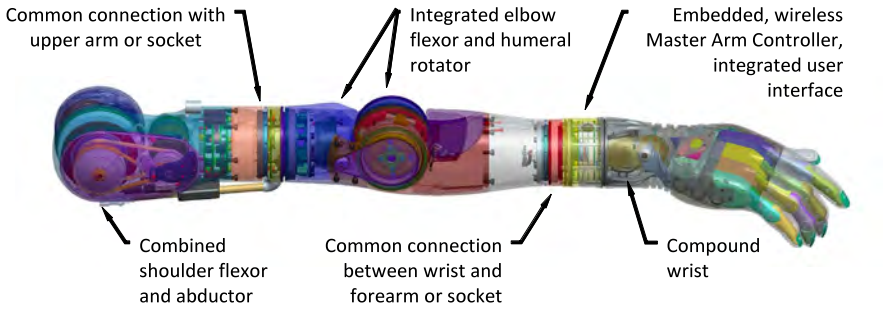
General features of the modular, fully configured arm (Figure 2) include 10 powered degrees of freedom (DOF), including shoulder abduction-adduction, shoulder flexion-extension, humeral rotation, elbow flexion-extension, wrist rotation, as well as a hybrid motion of wrist flexion-extension and radial-ulnar deviation.

Figure 2. Gen 3 arm shown in a configuration for scapulothoracic (ST) and shoulder disarticulation (SD) amputees (above). Translucent view with some of the features called out (below).

The arm system also includes 4 DOF associated with 6 distinct hand grips including chuck grip, power grip, tool grip, fine pinch open, fine pinch closed, and lateral "key" grip. The overall prosthesis and socket interface system is modular and capable of being configured for scapulothoracic (ST), shoulder disarticulation (SD), transhumeral (TH), and transradial (TR) levels of amputation [3].

Multiple control inputs are available for use as part of the arm system, including conventional techniques such as push switches, linear transducers, pressure transducers, and EMG sensors. The control system can also accommodate signals from more recent advancements in targeted reinnervation (TRI) and other central nervous system interfaces in development. A specific goal of the development process was to create a control scheme approach that allowed the prosthetist and the user to work together to choose control methods that are intuitive, effective, and appropriate for the specific situation of each user. Essentially, a toolkit is provided to support the prosthetist and therapist in control scheme development for each user.

With more mechanical degrees of freedom available within the prosthesis, additional control inputs were developed to support greater levels of simultaneous control and support increased usability. Our application of inertial measurement units (IMU) uses MEMS accelerometers and gyroscopes to provide additional DOF of translation and rotation signals and can be implemented at various locations on the body. Because there are typically limited sources of conventional signals for powered prosthesis control (usually 2 EMG, occasionally >2), conventional prosthetic devices are typically controlled in a serial fashion, i.e. from one joint to the next. With the Gen 3 arm possessing the capability of simultaneous powered multi-degree of freedom control and motion, these alternative control schemes allow greater simultaneous control for the user, even given limited EMG sites and without additional surgical intervention as would be required for other advanced or experimental control methods.



DESIGN FEATURES

Although numerous elements were important in the design effort, we will discuss two specific elements in more detail. They included 1) development of an efficient/effective means to control a full 10 DOF arm (with powered shoulder), and 2) the functional value of a wrist motion equivalent to ulnar/radial deviation found in the natural limb, a commonly requested articulation parameter by our users; this motion enables more efficient interaction of the prosthetic hand with objects on surfaces that are not at passive elbow height.

ENDPOINT CONTROL

Conventional control of a prosthesis by the user is often joint based, such as explicit command of supination of the wrist or flexion of the elbow. Typically these discrete motions are sequenced together into a series that eventually moves the terminal device of the prosthesis to its intended destination. More recently, certain motions have been bundled together, such as the simultaneous motion of fingers of the hand in an open and close maneuver.

The challenge with higher levels of amputation (ST, SD) is that they require a prosthesis with more degrees of freedom while also having a reduced number of sites to use as signal sources to control the prosthesis.

To address the limitations in control signals available by conventional means, we have instead implemented a method to control the position of the terminal device (hand) in space without primary regard by the user of the particular joint motion and/or sequence that is required to create the motion (Figure 3). The user simply indicates a movement of the endpoint (terminal device) forward/backward, up/down, right/left, or in combination, without needing to be concerned about how the shoulder, elbow or wrist joint needs to be articulated to achieve the ultimate destination.

The wireless IMU based sensors provide an excellent signal source for proportional control of the arm/hand endpoint motion in space. The software interface allows the system to be custom configured for the patient to define thresholds, velocity, and the configuration that is most intuitive to the subject. Thousands of hours of runtime have been logged with this control interface, it has been found to be extremely functional with minimal cognitive burden for the subject

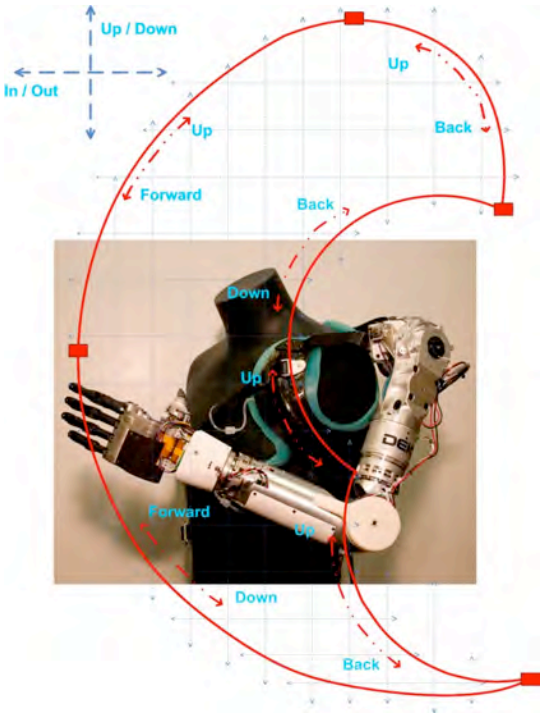


Figure 3. Endpoint control. Gen 2 arm with workspace boundaries.

WRIST ARTICULATION

The need for terminal device motion equivalent to ulnar and radial deviation has been a frequent request from therapists and patients during use of the arm system. This motion enables the subject to interact with and smoothly transfer objects to surfaces that are not at passive elbow elevation. For example, as shown in Figure 4, a bottle securely positioned in power grip requires radial deviation when the subject, in a standing position, plans to place the bottle in a stable perpendicular position on a table surface positioned below passive elbow level.

Figure 4. Examples of the hand in wrist extension-radial deviation position to place a bottle on a surface below the passive elbow position (left) and in a flexed, ulnar deviation position to place a bottle on an overhead shelf (right).

It is challenging to incorporate three independent DOF in the wrist with the lingering constraints of physical dimension, weight, and moment-arm costs. While the users and experts in the field made it clear that they valued ulnar/radial deviation; they also made it quite clear that they would not take that DOF if it eliminated supination and pronation or flexion and extension.

To assist us with understanding the spatial and temporal activity of the prosthetic arm's individual components, the arm incorporates logging features that allow tracking of positions, loads, and the power consumption of the joints during prosthetic arm activities. This allows the creation of a histogram profile of important parameters related to the various arm components during use. This quantitative data was essential as we progressed through the generations of prosthetic arm development. Based on this data and the observation of the prosthesis by subjects, a compound motion path combining wrist flexion/extension and ulnar-radial deviation was created that fulfilled the majority of wrist position functions required. Figure 5 illustrates the motion path of this hybrid degree of freedom. This allows the subject to access objects on surfaces well above and below the passive elbow position as noted in Figure 4 without requiring the cognitive burden associated with controlling these two DOF independently.

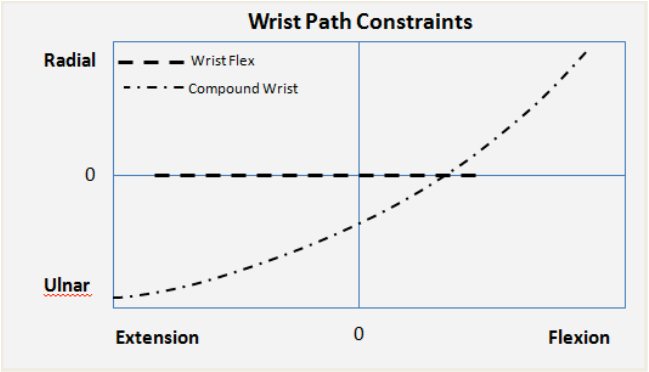


Figure 5. Example of motion curves for a compound (hybrid) wrist motion that incorporates wrist flexion-extension and ulnar-radial deviation

CONCLUSION

The collaboration of engineers, clinicians, and patients has allowed the development of an advanced upper arm prosthesis system that offers significant advances in functionality and capability; this development has required solutions to a variety of difficult design problems regarding arm capabilities, dynamics, and functionality as well as development of innovative control scheme components and improvements in interface design. The prosthesis system is proceeding through the final stages of development with continuing collaboration and feedback from user and prosthetist/therapist communities.

ACKNOWLEDGEMENTS

This project is sponsored by the Defense Advanced Research Projects Agency and the U.S. Army Research Office. Clinical studies were also sponsored by the US Department of Veterans Affairs. We would also like to acknowledge Next Step Orthotics and Prosthetics, Inc., biodesigns, Inc., and our research "test pilots" for their important and valuable contributions to this project. The information presented here does not necessarily reflect the position or policy of the government; no official endorsement should be inferred.

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CLINICAL EXPERIENCES WITH THE MICHELANGELO HAND, A FOUR-YEAR REVIEW

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ABSTRACT

With the integration of multiple grasp patterns, compliant hands have advanced the functionality of upper extremity prosthetic patients while also retaining the natural appearance of a human hand. The latest and most promising evolution in compliant hand technology is the Michelangelo hand by Otto Bock. A powered, opposable thumb is positioned electronically, smoothly transferring the hand into multiple grip patterns: lateral power grip, pinch grip, opposition power grip, tripod grip, finger abduction/adduction, full open palm and half open palm. Michelangelo operates significantly faster than previous compliant hands and includes a compliant flexion wrist that patients report has improved reliability and responsiveness. Enhanced software and EMG signal processing utilize an intuitive graphic user interface, promoting control predictability. In order to maximize the functional advantages of this technology, traditional occupational therapy training protocols should be modified to address multiple grasp function.

This presentation will examine the specific functional advantages of the Michelangelo hand based on four years of in-depth clinical involvement by this practitioner. The direct observations of 10 transradial level patients will be included, as will an overview of suggested modifications to occupational therapy training protocols.

THE I-LIMB PULSE HAND COMPARED TO THE I-LIMB AND DMC PLUS HAND

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INTRODUCTION

The new generation of multi articulating myoelectric prosthetic hands claims to be more functional than a one joint prosthetic hand. The aim of the study is to establish and compare the functionality of the myoelectric DMC plus, the i-LIMB and the i-LIMB Pulse hand.

CASE REPORT

In 2006, a 43-year-old man suffered from a wrist disarticulation at his dominant left side during work. Initially he was provided with a 2 electrodes myoelectric prosthesis with Dynamic Mode Control (DMC plus, OttoBock©). In December 2008 the patient received an i-LIMB hand (Touch Bionics©) with a ridged wrist and from December 2010 till May 2011 he used the i-LIMB Pulse with a friction wrist.

METHOD

The patient used different hands in a test procedure that covered all functional levels of the ICF. First we tested the DMC plus hand and after four weeks of usage the i-LIMB hand[1]. The i-LIMB was measured again after one year. The i-LIMB Pulse was tested after one month of training and daily use and again after three months.

Grip and pinch strength were measured using the Jamar dynamometer and Pinch meter of the e-LINK system.

Prehensile patterns and grip postures were assessed by the Southampton Hand Assessment Procedure (SHAP). A score of 98, ranging form 0-100, is proper for an unimpaired population.

The assessment of Capacity for Myoelectric control (ACMC 2.0) gauges myoelectric control in an everyday activity, packing a suitcase. A score of zero logits refers to an average control ability.

Satisfaction with the prosthesis was measured with the Trinity Amputation and Prosthesis Experience Scales (TAPES).

The functional status of The Orthotics and Prosthetics Users' Survey (OPUS) was established from a 19 item questionnaire. A score of 27 reflects zero logits and a moderate level of upper extremity function.

Visual Analogue Scale (VAS) scores were used to determine the patient's subjective opinion on strength, appearance, sound, precision grip, power grip, robustness and grip variety of the prosthetic hand. The patient scored also the relevance of these characteristics.

Finally in a semi structured interview, the patient told about his experiences with the prostheses.

RESULTS

Grip strength of the i-LIMB Pulse is almost equal to the strength of the DMC plus hand, and much higher than the grip power of the i-LIMB. The tripod grip strength is very much in favor of the DMC plus.

The Index of Function Score in the SHAP has improved for the i-LIMB from 52 after a month to a sore comparable to the DMC plus score [1]. The Pulse has the highest scores.

In the ACMC the patient has the highest score for the i-LIMB Pulse, and lowest for the i-LIMB.

The prosthesis satisfaction in the Trinity Amputation and Prosthesis Experience Scales is for the DMC lowest and highest for the Pulse. The adjustment to patients limitations is in favor of the i-LIMB Pulse.

The Functional Status in the OPUS is almost equal for the three tested hands.

According to the VAS scores the Pulse is highly valued for its variability in grip patterns, which is important to this patient. The DMC plus hand and i-LIMB Pulse both have a good grip power and are equally robust. The i-LIMB is the most vulnerable according to the patient's opinion.

In the interview the patient stated that what he liked best about the i-LIMB and i-LIMB Pulse compared to the DMC plus hand, was that he need not be very particular in positioning the i-LIMB hands before picking up every day objects such as a pen, a glass or a T-shirt, due to their fine precision grip.

DISCUSSION

This case report compares the functionality of the DMC plus, the i-LIMB and the i-LIMB Pulse hand.

In the first part of the study in which we compared the DMC plus and the i-LIMB [1], we suggested that the low scores for the i-LIMB hand in the SHAP might be due to the limited training, the extra time the thumb positioning took, the rigid wrist and the limited grip strength. The SHAP scores for the i-LIMB after one year improved to the level of the DMC hand. These suggest that more experience in using the i-LIMB hand improved the control of the hand and therefore took less time in performing the tasks. The i-LIMB Pulse has highest scores in the SHAP. The preset features of the Pulse in combination with the intensive daily training and the friction wrist seem to have contributed to the scores.

In the TAPES, the adjustment to limitation is highest for the i-LIMB Pulse. An explanation might be that the patient told that after the accident he had met new people, found new activities and goals in life which were directly related to his one handedness. He felt eventually that he had gained more in his life than he had lost. The adjustment to limitation might also be related to the lapse of time.

The high tripod grip force and power grip strength of the DMC plus, require high control ability when handling delicate objects. The Pulse has a comparable power grip, but a low tripod grip. This makes handling heavy and delicate objects possible.

CONCLUSION

Within the limitations of this casereport, we conclude, that the i-LIMB Pulse has a functional advantage over the i-LIMB hand. It has more power, is less vulnerable, and the functionality seems higher. The DMC hand is valued for its force and robustness, as is the i-LIMB Pulse. Training and every day use for at least four months is needed to be able to fit in a multi articulating myo electric prosthetic hand in daily activities. The preset features of the i-LIMB Pulse hand require intensive additional training to an experienced i-LIMB user.

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VA STUDY TO OPTIMIZE THE GEN 2 DEKA ARM: QUALITATIVE FINDINGS

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INTRODUCTION

In 2005 DARPA announced its "Revolutionizing Prosthetics" program and funded the development of the DEKA prosthetic arm. When the Gen 2 prototype DEKA Arm System became available for clinical research and testing, DARPA signed a Memorandum of Agreement with the Department of Veterans Affairs (VA) and provided additional funds to DEKA to support a VA Optimization Study of the DEKA Arm system. The purpose of this study was to obtain user feedback to inform design of the next prototype, the Gen 3 Arm.

The DEKA Arm is designed for users with amputations at the forequarter, shoulder disarticulation, transhumeral or transradial level. There are three versions available: shoulder configuration (SC), humeral configuration (HC) and radial configuration (RC). The Gen 2 Arms that we tested had 6 hand grips (power, chuck, lateral pinch, open pinch, closed pinch and tool grip), and used a variety of control inputs including EMGs, air bladders, and foot controls (Force Sensitive Resistors [FSR]) and Inertial Measurement Units [IMUs]).

STUDY DESIGN AND PURPOSE

The VA study was an iterative usability and optimization study employing a multiple case study design with a mixed-methodology approach. Concurrent quantitative metrics and qualitative data were collected to provide richer, more valid, and more reliable findings than a design based on either the qualitative or the quantitative method alone. The purpose of this paper is to describe our VA subjects' perspectives on using the Gen 2 DEKA Arm.

METHODS

Subjects

Twenty-six subjects were fit with the DEKA Arm (22 men and 4 women), ages 19 to 82 years. Five were on active duty in the U.S. military, 13 were veterans (not on active duty), and 8 had never served in the U.S. Armed Forces. Twenty-three subjects had unilateral upper arm loss and 3 had bilateral upper limb loss. Ten subjects were fit with SC DEKA Arms, 8 were fit with an RC, and 8 were fit with an HC. Four of the 10 subjects fit with a SC had short transhumeral amputations. Subjects were seen at one of four participating sites, VA NYHHS (Manhattan), James Haley VA (Tampa), Long Beach VA (Long Beach), and the Center for the Intrepid (CFI).

Data Collection

Subjects were told that the primary objective of the study was to obtain feedback on the DEKA Arm prototype in order to inform the design efforts of the next prototype, the Gen 3 Arm. Their opinions about all aspects of the DEKA Arm were solicited throughout the study through surveys, semi-guided interviews, audio memos, and videotaped training and testing sessions. Subjects had approximately 20 hours of training in the use of the DEKA Arm (some subjects with SC configuration had up to 30 hours of training) and participated in multiple testing sessions.

Data Analysis

Qualitative data analysis involved open coding of transcripts from audiotapes, memos from videotaped sessions, and participants' responses to open-ended survey questions. Open coding was used to reduce the data to a set of important themes or categories. The data was synthesized in a cross-group analysis to compare similarities and differences in experience and recommendations of participants by DEKA Arm level. At each stage of data collection and analysis, members of the research team discussed key case findings and interpretations.

RESULTS

Main Impressions

At the end of the study subjects were asked, "What is your impression of the DEKA Arm?" A majority of subjects (21/25, 84%) had favorable impressions; 14 of these as unequivocally favorable and 7 were favorable with critical feedback. A higher percentage of subjects using the SC Arm were classified as unequivocally favorable (70% of SCs; 43% TC; 50% RC). The 7 subjects who had generally favorable impressions but had critical feedback commented on issues,

including weight, reliability, ROM of the wrist, and the need to put the arm in standby mode while walking. Four subjects (16%), including one subject at each level, had unfavorable impressions of the DEKA Arm at the end of the study. Weight was the most commonly cited criticism, but it was only one among a variety of issues mentioned.

Function with the DEKA Arm

Twenty-one of the 22 subjects who used a prosthesis prior to the study gave examples of new activities they had been able to perform with the DEKA Arm during the training protocol that they had not been able to perform with their current prostheses. The most frequently mentioned types of new functional activities were self-care and everyday household/office tasks. At the end of the study subjects were asked if there were any activities they could not do with the DEKA Arm that they were able to do with their current prostheses. Among the 22 subjects who answered this question, 77% said "no" while 23% answered "yes." Examples of activities from those who answered "yes" included: wash myself, drive a car, ride a bike. Some of these tasks were obviously related to limitations of the foot controls and the level of water resistance of the Gen 2 prototype.

Desire to Receive a DEKA Arm in the Future

At the end of the study subjects were asked if they would want to receive a DEKA Arm in the future and to explain why or why not. Nineteen out of 25 subjects (76%) answered "Yes." Many subjects explained that they wanted a DEKA Arm because of increased overall function, saying, for example, "will make everyday activities better", "would open up a whole new world of independence and quality of life." Eighty percent of those using an SC Arm, 86% of HC Arms, and 63% of RC Arms clearly wanted a DEKA Arm in the future. Two subjects stated definitively they did not want the Arm in the future (1 SC, 1 HC), while 4 said "Maybe" (3 RC, 1 SC). Among the subjects who said "Maybe", all that were users of the RC listed weight as a reason they may not want the Arm system.

Feedback on Grips

At the end of the study subjects were asked which, if any, of the hand grips they found most useful. Open and closed pinches were most frequently mentioned as most useful, followed by lateral pinch and chuck. At the same time subjects were also asked if there were any grips they would not use: 64% stated there were no grips they would not use and 36% thought there were one or more grips they would not use. Tool grip was most listed most frequently, followed by chuck grip.

Concerns about taking the DEKA Arm home

Subjects were asked to list their concerns about using a Gen 2 DEKA Arm at home. Twenty-one out of 25 subjects who responded to this question (84%) expressed at least one concern about using the prototype DEKA Arm at home, largely relating to repairs, water resistance, and weight. Despite concerns, 76% said they wanted to receive a DEKA Arm in the future. Subjects provided feedback on features of the arm system such as cosmesis, weight, controls,donning and doffing.

Recommended Improvements

Subjects were asked, "How do you think the DEKA Arm system could be improved to make it easier to use and more acceptable to other persons with upper limb loss?" The most frequently mentioned improvements were 1) to make it lighter in weight and 2) to improve controls. The next three most frequently mentioned categories related to 1) making the Arm system smaller/wireless/with less external components; 2) improving the wrist motions and 3) improving the socket fit and/or ease of donning and doffing the Arm system. Six subjects suggested improvements in cosmesis. Other less frequently mentioned improvements related to reducing noise, and improvements to the inflatable bladders or inflation process for socket bladders, made by 3 HC subjects.

CONCLUSION

Our subject's overall impressions of the Gen 2 DEKA Arm were favorable. The majority of subjects expressed a desire to obtain DEKA Arms in the future. Many of these same subjects also expressed critical feedback about the Arm, but this fact alone did not equate with desire to receive a DEKA Arm in the future. All subject feedback was shared with DEKA and many suggested improvements have been addressed in the Gen 3 design.Clinical studies of the Gen 3 Arm are now underway.

ACKNOWLEDGEMENTS

This research was supported by VA RR&D, VA RR&D A6780 and VA RR&D A6780I. DEKA's support of the VA optimization studies was sponsored by the Defense Advanced Research Projects Agency and the U.S. Army Research Office. The information in this manuscript does not necessarily reflect the position or policy of the government; no official endorsement should be inferred.

The authors acknowledge the valuable work of study analyst Shana Lieberman Klinger and study coordinator Kate Barnabe.

USING THE CONTROLLER AREA NETWORK FOR COMMUNICATION BETWEEN PROSTHESIS SENSORS AND CONTROL SYSTEMS

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ABSTRACT

Advancements in electronics technology have yielded a wide variety of low-cost intelligent sensors and microprocessor-based control systems that can be incorporated into prostheses. The most efficient and reliable way to interconnect these components is through a communication bus. We have used the Prosthetic Device Communication Protocol (PDCP), based on the Controller Area Network (CAN), to develop a distributed control system for both upper and lower limb prostheses. The upper limb implementation is comprised of a master controller, a 16-channel myoelectric signal (MES) data acquisition (DAQ) module, and a modified Boston Digital Elbow (Liberating Technologies, Inc.). The lower limb implementation is comprised of a master controller, a 12-channel MES DAQ module, a 16-channel electromechanical sensor acquisition module, and an experimental knee prosthesis manufactured by Vanderbilt University. Bus utilization was, on average, 62% for the upper limb system and 73% for the lower limb system with no loss of data or perceivable latency. This contribution highlights the suitability of the CAN bus to support the data transfer required to control powered prostheses. It also supports the PDCP protocol as a higher-level protocol to facilitate interoperability between prosthesis control systems.

INTRODUCTION

Upper limb myoelectric prostheses are in widespread clinical use, and a number of different companies manufacture and sell a variety of components. Furthermore, many research groups are developing new components including, but not limited to, smart electrodes, anthropomorphic terminal devices, sensory feedback devices, and actuated knees and ankles. These devices need to communicate if they are to be used together in a functioning prosthetic system.

Meetings were held between researchers and vendors at the 2005 and 2008 Myoelectric Controls Symposiums and at the 2007 and 2010 International Society for Prosthetics and Orthotics Conferences to discuss the creation of an open, standardized communication bus. The University of New Brunswick led the development effort and recently released a draft of a bus protocol that uses a Controller Area Network (CAN) bus. The protocol is called the Prosthetic Device Communication Protocol (PDCP) [1]. A CAN-based communication method allows the number of wires necessary for data transmission to be dropped from at least two wires per sensor to a total of four wires for the entire system; specifically, two for the differential communication pair, one for power, and one for ground. CAN supports data transfer rates of up to 1 megabit per second and operates robustly in noisy electronic environments. In addition, a number of small, low-power, cost effective, and reliable CAN-capable microcontrollers have been developed for use in embedded systems and in the automotive industry.

The PDCP is a master/slave protocol that utilizes a unique 8-bit identifier for each electronics module present on the CAN bus. Every message sent across the bus is comprised of the identifier followed by a data payload of up to 8 bytes. This identifier acts as an 'address' and allows the master to communicate with a specific module or allows a module to communicate with the master.

The PCPD has been implemented on both upper and lower limb prosthetic components at the Center for Bionic Medicine (CBM) at the Rehabilitation Institute of Chicago. At CBM, we fit powered upper and lower limb prosthesis prototypes to amputee patients and have been working on development of advanced embedded controllers for both upper and lower limb prostheses. Because many of the same electronics system components can be used to support both upper and lower limb devices, we implemented the PDCP protocol to simplify the interoperability between our modular electronics devices.

SYSTEM PROFILES

Lower Limb System

Myoelectric control for lower limb prostheses is an emerging field of research and development. Microprocessor-controlled variable damping knees and powered knee/ankles use state-based control to interpret signals measured from physical sensors embedded in or attached to the prosthesis.

It has been shown that MESs may be interpreted by pattern recognition algorithms to decode information regarding the mode of activity that the patient is attempting to perform [2]. Preliminary experiments completed to determine if MESs added useful control information were accomplished using an instrumented passive prosthesis (Fig. 1a) with conventional wiring. We have modified a powered knee prosthesis

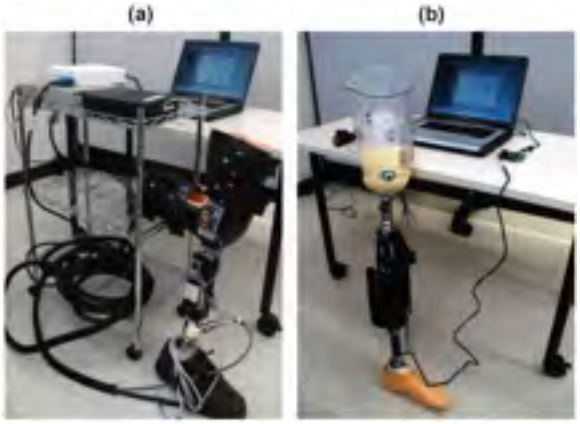

Figure 1: Data acquisition and control systems showing (a) an instrumented passive prosthesis connected to a laptop using conventional wiring and (b) a powered knee prosthesis connected to a laptop using a CAN bus.

manufactured by Vanderbilt University (VU) to use the PCDP to stream sensor information to, and accept commands from, a master control module. Thus, the overall system (Figure 1b) is now comprised of an MES DAQ module, a physical sensor module, the VU powered knee module, and a master control module that runs Control Algorithms for Prosthetic Systems (CAPS) software [3]. CAN communication is used to connect the modules, with conversion between CAN communication and (1) the computer's Universal Serial Bus (USB) input and (2) the VU powered knee's Serial Peripheral Interface (SPI) input (Figure 2).

MES DAQ Module

The MES DAQ module is comprised of 12 precision instrumentation amplifiers (Analog Devices AD8295) providing a gain of 50V/V as well as a second-order high-pass filter at 26 Hz. A second signal conditioning stage is made up of a second-order multiple feedback band-pass filter (Analog Devices AD8626) with a gain of 20V/V, a center frequency of 126 Hz, a low-frequency cutoff of 26 Hz, and a high-frequency cutoff of 387 Hz. The outputs of the instrumentation amplifiers are sampled by two 8-channel 16-bit analog-to-digital converters (ADCs) (LTC1859, Linear Technology) at a default sampling rate of 1 kHz. The digital outputs from the ADCs are collected by a PIC32MX795 32-bit microcontroller (Microchip Technology), which has been programmed to implement the PDCP.

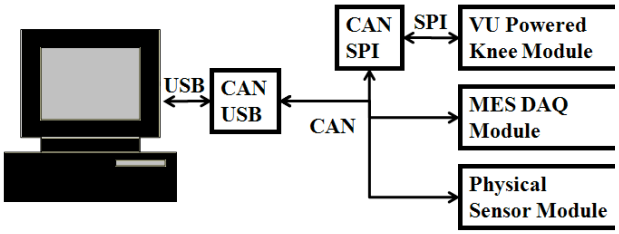

Figure 2: Lower limb distributed control system

The MES DAQ module was assigned a unique system identifier, 0h50. A graphical user interface was created within CAPS to allow the user to specify parameters associated with the MES DAQ devices such as the sample rate, bit resolution (up to 16 bits per sample), software gain, and number of channels to be streamed across the CAN bus. The CAN bus allows a payload of at most 8 bytes of data to be sent within each message. Consequently, the data has been packaged such that samples from up to four channels are sent with each CAN message. Each data package uses the first byte as a sequence counter, so that statistics regarding missing data on the CAN bus can be tracked by CAPS and used for debugging purposes.

Physical Sensor Module

The physical sensor module was constructed at CBM to provide bridge excitation voltage to a six degree of freedom MCW-6-1K load cell (AMTI) and amplify the resulting strain gauge measurements. The module also houses a six degree of freedom inertial measurement unit manufactured by SparkFun Electronics (SEN-09431). Finally, four additional analog sensors, such as force sensitive resistors or potentiometers, may also be supported depending on the experimenter's needs. All signals are sampled by two 8-channel, 16-bit ADCs (LTC1859) at a default sampling rate of 500 Hz. The digital output from the ADCs are collected by a PIC32MX795 32-bit microcontroller (Microchip Technology, Inc.) that has been programmed to implement the PDCP. All digital signals are input to the PIC32 which has been programmed to implement the PDCP.

The physical sensor measurement module was assigned a unique system identifier, 0h40. The same parameters as described for the MES DAQ unit can be configured using the CAPS graphical user interface and the data is packaged in a similar manner.

VU Powered Knee Module

The powered knee prosthesis is equipped with a low-level control system with expected impedance parameters virtual stiffness, k_k, virtual equilibrium angle, ,θ_{ek}, and virtual

damping coefficient, b_k, as inputs so that a knee torque command, τ_k, can be created according to equation 1:

$$\tau_k = k_k(\theta_k - \theta_{ek}) + b_k\dot{\theta}_k \quad (1)$$

where the subscript k indicates that the parameters or measurements are associated with the knee, $\theta_{k\theta_k}$ is the knee angle measurement from the prosthesis, and $\dot{\theta}_{k\dot{\theta}_k}$ is knee velocity. All of the above impedance parameters are sent to the knee from CAPS over the CAN bus.

The VU powered knee prosthesis also contains several intrinsic physical sensors. The physical sensor data include torque applied at the knee motor (in Nm), angular velocity (in degrees/s), and the angle at the knee joint (in degrees).

The VU powered knee module was assigned a unique system identifier, 0h20. The knee impedance parameters are packed into a single CAN message that is sent at a default sampling rate of 40 Hz at 16-bit resolution. The knee physical sensor data are transmitted over the CAN bus with 16 bits of resolution at a frequency of 500 Hz

CAN-SPI Bridge Module

The VU powered knee physical sensor data and the knee impedance parameters are transmitted to and from the CAN bus via an SPI-to-CAN bridge module. This module, developed at CBM with the assistance of VU, packages the knee physical sensor information collected via the SPI bus into a CAN message compatible with the PDCP. Conversely, the CAN-SPI bridge unpacks the impedance parameters from the PDCP CAN messages and transmits those parameters to the knee intrinsic control system via SPI. A single PIC32MX795 microcontroller handles all of this data translation.

CAN-USB Bridge Module

The CAN-USB bridge module performs a similar function to the CAN-SPI module, but uses a PC-friendly USB protocol as opposed to an SPI protocol. The CAN-USB module is used to unpack CAN messages for transmission via USB to the PC for processing by CAPS. Knee control parameters are sent to this module via USB, packed into a PDCP compatible CAN message, and then transmitted over the CAN bus. This module, developed at CBM, consists of a single PIC32MX795 microcontroller that interprets the data in both directions.

Upper Limb System

The use of MES-controlled upper limb prostheses is commonplace at CBM. We continuously strive to improve our control system electronics, data collection methods, pattern recognition algorithms, system robustness, and ease of use for the patient. To improve upon wire management and interoperability for our upper limb systems, we have implemented the PDCP and a CAN bus on them as well. The upper limb control system consists of an MES DAQ module, LTI Boston Elbow, and PC, all connected via CAN bus, with conversion from CAN to analog signals for the Boston Elbow (Figure 3).

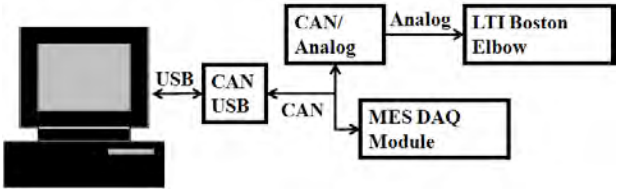

Figure 3: Upper limb distributed control system.

MES DAQ Module

The analog front-end of the MES DAQ module for the upper limb system is very different from our lower limb system due to the space constraints encountered with upper limb systems. The most distal level of amputation for an upper limb amputee considered here is wrist disarticulation, which requires mechanical and electronic packaging size to be no larger in diameter than that of a wrist. Using the wrist diameter of the LTI Boston Elbow as a size constraint, CBM implemented a single integrated circuit (IC) designed for biopotential data acquisition. This 8 x 8 mm IC (RHA2216, Intan Technologies) provides 16 fully differential input amplifiers, an internal bandpass filter, a mid-range gain of 200V/V, and an SPI for use with microcontrollers. For our application, the first order high-pass filter was set at 30 Hz and the third-order low-pass filter was set at 452 Hz. The multiplexed output of the IC is sampled by a single-channel, 16-bit ADC (AD7680, Analog Devices) at a rate of 1 kHz. A smaller 16-bit microcontroller (PIC24HJ128GP502, Microchip Technology) was used to collect the digital output of the ADC and transmit the data over the CAN bus according to the PDCP.

As was the case for the lower limb MES DAQ, the upper limb MES DAQ used the identifier 0h50. This allowed channel configuration via CAPS. If both an upper and lower limb MES DAQ module were present in the same system, they would be assigned separate identifier values.

LTI Boston Elbow

The upper limb device used in this system was a four degree of freedom (DOF) LTI Boston Elbow (Liberating Technologies, Inc.). This device was set up to use antagonistic analog input voltages in the range of 0–5 V to control each degree of freedom. The Boston Elbow input was modified

so that CBM could input analog signals that were output commands from pattern recognition algorithms performed within CAPS. The analog voltage outputs to the Boston Elbow device represent percent of DOF speed, between 0% and 100%, corresponding to a 0–5 V Boston Elbow input.

CAN-Analog Bridge Module

This module consists of a microcontroller (PIC32MX795) that translates incoming CAN messages into outgoing analog voltage signals that simulate EMG signals in order to properly control the Boston elbow. A digital-to-analog converter (DAC) (MAX5307, Maxim Integrated Products) provides up to eight channels of simulated EMG to drive the four DOFs. The microcontroller provides the means of translating the control signals from CAPS, in the form of a CAN message, to the 0–5 V analog signals recognized by the Boston Elbow.

RESULTS

The CAN-based PDCP was implemented in both lower and upper limb systems within our laboratory and used for real-time data acquisition and control of powered prostheses. Bus utilization is the percentage of bus bandwidth used during data transfer; high bus utilization means heavy traffic on the bus. Bus utilization is an important metric to consider for communication system designers: as the bus data transfer rate reaches its theoretical capacity, new information can no longer be transmitted due to lack of available bandwidth. Equation 2 is used to calculate bus utilization (bU):

$$bU = \frac{(nMsg * AvgBit) - (SBy * BStf)}{(Bcap) * \tau} \quad (2)$$

where nMsg is the total number of CAN messages received or transmitted by CAPS, AvgBit is the average number of bits per CAN message with a full data payload of 8 bytes (120.8 bits [4]), SBy is the cumulative number of bytes less than the full payload, BStf is the average number of bits possible per byte including the stuff bit (8.8 bits), Bcap is the CAN bus bit rate capacity of 1Mbit/s, and τ is the total elapsed time in seconds.

Bus utilization with just the VU Powered Knee (6%) increased by 24% when the CBM physical sensor module was added to the bus and by an additional 33% when the MES DAQ module was added to the bus (Table I).

In the upper limb system, bus utilization was 61% with the MES DAQ module but increased slightly to 62% when the CAN-Analog bridge module was added (Table II).

Table I: Bus Utilization for Lower Limb System

Module Added	# of Channels	Resolution	Sample/ Update Rate	Bus Utilizations
VU Powered Knee	3 Inputs 3 Outputs	16 bits 16 bits	500 Hz 40 Hz	6%
CBM Physical Sensor	16	12 bits	500 Hz	30%
MES DAQ	12	16 bits	1 kHz	73%

*Includes operating VU Powered Knee Module
**Includes operating VU Powered Knee Module and CBM Physical Sensor Module

Table II: Buss Utilization for Supper Limb System

Module Added	# of Channels	Resolution	Sample/ Update Rate	Bus Utilizations
MES DAQ	16	16 bits	1 kHz	61%
CAN Analog	1	16 bits	40 Hz	62%

*Includes operating MES DAQ Module

DISCUSSION

The CAN bus was a clear choice for this project due to a few major design goals: reduction in the amount of wiring needed for data acquisition and control, high data transfer rate, robust communication in electrically noisy environments, low cost, commercially available hardware, and widely available firmware application programming interface support.

CBM also recognized the effort of almost a decade of research and development of a bus-based communication standard for powered prostheses [5-7]. An industry standard for system communication would facilitate interoperability between devices and may allow the patient greater flexibility in choosing prosthetic components.

The results show that a large majority of the CAN bus may be utilized without data lost due to bus bandwidth restrictions. With an average bus utilization of 73%, a system designer can still transmit more data if needed. In other studies, CBM has utilized up to an estimated 93% of bus capacity before seeing data dropped due to bandwidth limits.

CONCLUSION

We have shown that the use of the CAN bus with the PDCP is sufficient to support the bandwidth necessary for real-time control and data acquisition of both lower and upper limb powered prostheses in distributed control systems. Our hope is to continue to build upon our knowledge of the CAN bus, to continue to support and assist in the development of the PDCP, and to strive for an industry-accepted communication protocol.

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BRIDGING THE GAP: ENSURING COMMUNICATION BUS STANDARD COMPATIBILITY WITH CURRENT COMMERCIALLY AVAILABLE PROSTHETIC COMPONENTS

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ABSTRACT

The creation and adoption of an open digital communication standard has long been identified as a necessary progression for the prosthetic field. Although there are many obstacles facing such an effort, maintaining compatibility between future devices and currently available hardware is one of the first obstacles that must be overcome. This paper will highlight one method of addressing this issue.

BACKGROUND

The development of an open and standardised communication platform must successfully overcome several challenges prior to becoming a viable option for the prosthetic industry. Obstacles to the implementation of any standard can vary from one company/institution to another. Limited development resources and funding are prime examples of the challenges facing this type of initiative. Widespread adoption will also be strongly affected by the protocol's flexibility in various applications, its capability to improve/augment the current standard in the industry, and its ability to support commercially available prosthetic components and devices. This paper will address some of these issues and illustrate some solutions through the use of the University of New Brunswick's (UNB) Prosthetic Device Communication Protocol (PDCP).

COMMUNICATION BUS OVERVIEW

The electrical standard in the prosthetics industry has evolved over the years to use 0-5V analog voltage levels to pass information between components. As prosthetic systems have become more complex, it has become necessary to pass much more information between the various components of the system than can be supported by one or two analog voltage signals. As a result, several research systems have incorporated a serial bus-based communication between the various components [1,2]. Otto Bock has become the first manufacturer to use a serial bus in a commercially available system with the introduction of the Axon Bus in 2005 [3].

The creation of an open digital standard for intercomponent communication in the prosthetic field has been a much-discussed topic for the past several years [4-6] and has been the focus of meetings and workshops in past MEC Symposiums and ISPO World Congresses. Through these efforts, the Standardised Communication Interface for Prosthetics (SCIP) group was formed in an effort to develop such a standard. Seeing a potential to advance the SCIP initiative, UNB offered to share technical details of their PDCP system as they became available.

The PDCP is a digital serial communication bus based on the Controller Area Network (CAN). It was created within the framework of the UNB Hand System project [7] in an attempt to provide a reliable and robust communication platform between the various system components. Efforts have been made to ensure that the protocol would be flexible in order to support future bandwidth and low power consumption requirements as the project progressed.

METHODOLOGY

The PDCP system has been developed to support modularity of components in the UNB Hand System, but also with the intention of becoming the starting point of an industry standard. In order to gain acceptance in the industry, the need to support existing components has been recognized.

Through the course of the PDCP development, it was quickly identified that the lack of support for currently available devices and sensors would be a major limitation of the standard, if not addressed. As with the introduction of any new system, the system must be phased in; introducing new features while supporting the well established and accepted features of the existing system. This is especially true in the slowly evolving prosthetics industry.

To enable support for existing components, a device has been developed (Figure 1) to translate between the analog signals common to most current systems and the signals expected on the PDCP bus. This PDCP translator hardware allows up to four analog inputs within the 0-5V range to be brought into a PDCP-based system or four analog 0-5V outputs

to be generated from a PDCP-based system. Therefore, any combinations of up to 4 sensors can be connected to a PDCP-based system with the use of a single PDCP translator circuit board. These sensors may include: EMG electrodes, force sensitive resistors (FSR), linear potentiometers, switches, etc. In addition, any combinations of up to 4 actuators can also be connected to a PDCP based system with the use of a single PDCP translator circuit board. These actuators may include: hands, wrists, elbows, etc. If more than 4 sensors or more than 4 actuator outputs are required, additional PDCP translator boards may simply be added to the system.

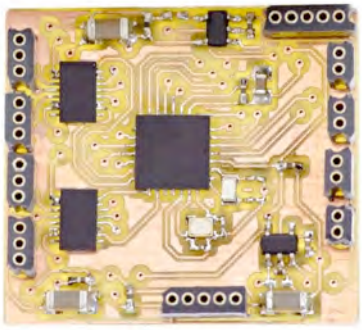

Figure 1: PDCP Translator Board

This approach allows new prosthetic systems to implement and take advantage of the PDCP bus while supporting existing sensors and actuators. As such, a complete system could consist of purely analog sensors and actuators from existing commercially available components but use a serial bus based interconnect incorporating the PDCP. One benefit of such an implementation would be the reduced number of wires running through the prosthetic system. The interconnection of all the sensors and actuators would be reduced to the four wires needed for the PDCP bus as illustrated in Figure 2.

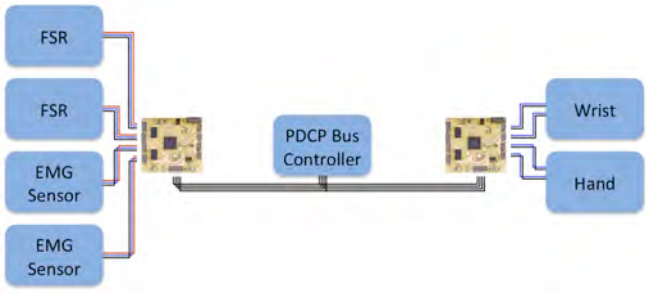

Figure 2: Example Using Existing Components within a PDCP-Based System

CONCLUSION

The development of an open communication standard must include a mechanism to support existing components that are currently used in the prosthetic industry. In terms of the PDCP standard, the development and implementation of the PDCP translator hardware has satisfied this requirement by allowing the integration of commercially available prosthetic components with no modifications required to these off-the-shelf devices.

ACKNOWLEDGMENTS

This project is sponsored by the Atlantic Innovation Fund, the Natural Sciences and Engineering Research Council of Canada, and the New Brunswick Innovation Foundation. Additional funding has been provided by the University of New Brunswick and Liberating Technologies Inc.

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TOWARDS A UNIVERSAL COUPLER DESIGN FOR MODERN POWERED PROSTHESES

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INTRODUCTION

Universal coupler standards have played a critical role in allowing prosthetists to choose the best terminal device for their patient's particular needs. The conventional quick-disconnect coupler allows users to passively rotate their prosthesis or to use a weak motorized rotator. However, the current standard precludes use of the strong motorized wrist rotators introduced by several companies and universities, as these devices would decouple the current universal coupler. A new universal coupler standard is required to allow interchangeability of these new devices. An open source universal coupler standard that meets necessary design requirements would better serve prosthesis users.

DESIGN REQUIREMENTS

We have identified the following eight major considerations that the design of an ideal universal coupler should meet in order to accommodate the needs of the prosthesis user and prosthetist:

1. Size constraints:
    a. Axial length should be 12 mm or less
    b. Outer diameter should be 25 mm or less
2. Ease of connection and disconnection:
    a. If multiple actions are required, they should be sequential
    b. Should allow prosthesis to be donned and doffed with cosmesis covering in place
3. Rotation constraint:
    a. Device should lock to prevent unwanted rotation
    b. Number of mechanical locking orientations should equal the number of permissible electrical connection orientations
4. Strength constraint:
    a. Device should withstand 45 Nm of bending torque (70 Nm desired)
    b. Device should withstand up to 15 Nm of axial torque
5. Interface with adjacent segment:
    Adjacent segment may be larger than coupler diameter
    a. Contact with an adjacent segment larger than the coupler may increase connection strength
6. Manufacturing constraints:
    a. Proximal component may be a stand-alone part or made integral with a lamination collar
    b. Distal component may be a stand-alone part or made integral with the distal device
    c. May be manufactured from different metals depending on strength requirements of user
7. Location:
    a. May be used at the wrist
    b. May be placed near, but slightly distal to, the elbow for modular forearms
8. Electronic constraints:
    a. Male and female connectors for six conductors
    b. Should prohibit improper electrical connection
    c. Should preclude the possibility of electrode shorts during connection or disconnection

Size constraints

The size of the coupler is important for both user function and cosmetic appearance. Patients with long residual limbs will benefit from a coupler with a short axial profile, 12 mm or less, as it will not add significant length to the limb. The coupler diameter must be no larger than the minimum dimension of a small wrist, 25 mm, so that with a cosmesis covering, it will have the appearance of an anatomic wrist. The current conventional quick-disconnect coupler has a length of approximately 18 mm and a 40 mm diameter.

Ease of connection and disconnection

Connecting and disconnecting the coupler should require minimal dexterity so that it is manageable for all users, including bilateral amputees. For this reason, connecting and disconnecting the coupler should require as few user actions as possible, and if multiple actions are required, they should be sequential rather than simultaneous. Many cosmetic coverings extend from the hand to the elbow and would therefore prevent direct access to the coupler. The coupler must be easy to connect and disconnect without visual feedback and without requiring removal of the cosmetic covering.

Rotation constraint

In order for the coupler to be fully functional, it must lock into position and not rotate passively in the presence of pronation or supination forces. The number of possible rotational positions for mechanical connection of the coupler should take into account the necessity of preventing incorrect electrical contact.

Strength constraint

We propose the coupler must withstand a minimum 45 Nm of bending torque and 15 Nm of axial torque based on previous studies of maximum torque in able-bodied persons [1]. However the desired coupler strength is 70 Nm in bending based on the maximum passive resistance in elbow designs [2]. Ideally, the coupler would be manufactured from a material strong enough for all users; however, various material grades may be required to meet the extreme force requirements of some users.

Interface with adjacent segment

For a proper anatomic appearance, it may be desirable for the segment just distal or proximal to the coupler to be larger than 25mm. Therefore, the coupler must not hinder a larger adjacent segment. In instances where the user's adjacent body is larger than the coupler, the strength of the connection may increase.

Manufacturing constraints

The design of the coupler must allow it to be manufactured as an integral part of an adjacent segment such as a laminate forearm or terminal device, or as a stand-alone product that may be attached to an adjacent segment with screws or other means.

Location

Due to the variable location of amputation sites among users, as well as differences in design of current and future prostheses, the coupler should be capable of being located at any position along the length of the forearm.

Electronic constraints

Electrical power and communication signals must be able to pass through the coupler. Electrical connections must remain functional through frequent mechanical and electrical connection/disconnection cycles. The device must contain enough contacts for all necessary electrical communication: we have determined that six electrodes are sufficient for electrical power and communication needs. Careful design of the coupler is necessary to prevent improper electrical connections or electrical shorts during connection and disconnection. Furthermore, when coupled the coupler design should prevent or limit exposure of electrical connections to moisture.

CONCEPTS

The three universal coupler concepts presented herein illustrate possible mechanical and electrical features that have been developed to address these eight design considerations. These designs were independently developed, yet all three have striking similarities. Each concept uses multiple tabs to transfer forces across the two components and engagement of the coupler requires a sequential translation followed by rotation.

Concept 1

Concept 1 (Figure 1) has an axial length of 10 mm and a 25 mm diameter. The main features of the proximal (light tan) and distal (dark teal) components are shown in Figure 2. Component assembly is shown in Figure 3.

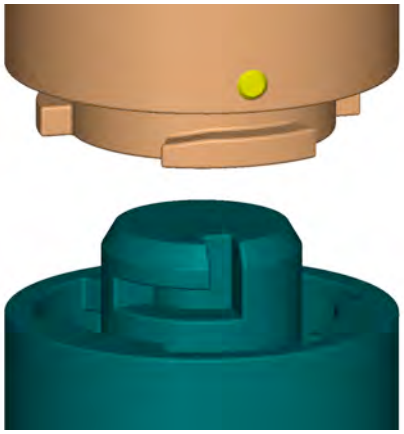

Figure 1: Universal Coupler, Concept 1

Care has been taken to ensure that these components may be assembled without getting hung-up during assembly. As the two components are brought together, the chamfered edge (E, Figure 2) guides the male shaft of the distal component (D, Figure 2) into the hollow center of the proximal component.

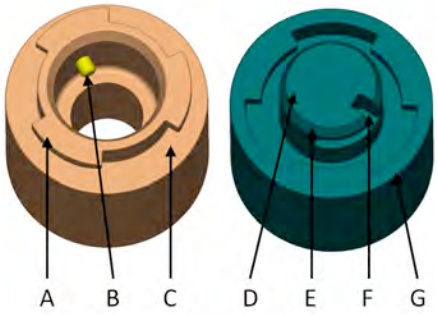

Figure 2: Features of Universal Coupler, Concept 1. The proximal component is shown on the left (light tan), and the distal component on the right (dark teal).

The proximal and distal components slide together for 5 mm until the transverse pin (B, Figure 2) on the distal component contacts the male shaft (Figure 3b). The distal component must then be rotated until the keyed shaft (F, Figure 2) is aligned with the transverse pin. This feature creates one unique orientation for coupler assembly. Once the transverse pin and keyed shaft are aligned, the distal component translates another 5 mm until the outer rims of both components make contact (Figure 3c). Three large tabs (A, Figure 2) provide strength and resistance to bending. Locking the coupler requires a 60° rotation of the distal connector (Figure 3d). The locking switch and electronic contacts have been omitted from these figures.

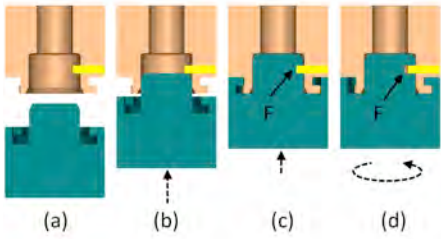

Figure 3: Assembly of Universal Coupler, Concept 1: (a) Unassembled, (b) Partial assembly, (c) Full assembly, (d) Full assembly after lock. Distal component is shown in dark teal, proximal component in light tan.

Concept 2

Concept 2 (Figure 4) has an axial length of 9.5 mm and a 25 mm diameter. It is based around an auto-locking system and is composed of four components. The proximal component (black) and the distal component (green) form the structure of the wrist, with the latch (red) and return spring (silver) providing the locking mechanism (Figure 5).

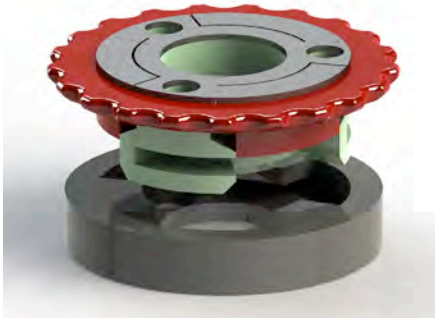

Figure 4: Universal Coupler, Concept 2

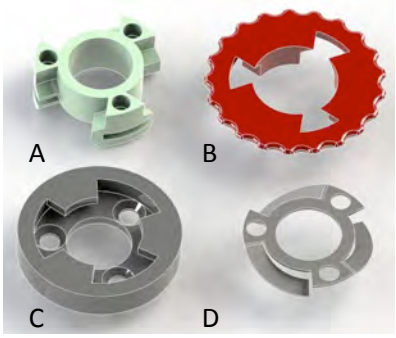

Figure 5: Individual Components of Concept 2: (A) distal component (green), (B) latch (red), (C) proximal component (black), (D) return spring (silver)

As the distal and proximal units come into contact, chamfered edges (B, Figure 6; Figure 7A) align the tri-leaflet pattern (D, Figure 6) to its corresponding cavity (C, Figure 6) and initiate the depresion of the latch.

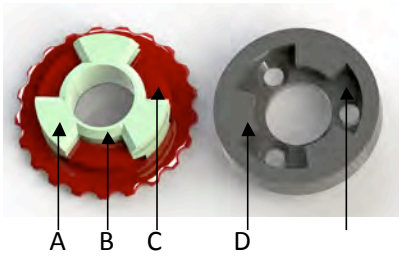

Figure 6: Features of the Universal Coupler, Concept 2. The distal component is shown on the left (red & green) and the proximal component on the right (black).

As the distal unit is inserted, the leaflets continue to depress the latch; once this reaches a maximum axial depth of 3 mm within the proximal unit (E, Figure 6; Figure 7B), it is rotated counter-clockwise by 60 degrees by which time the leaflets are fully engaged within their corresponding cavities (A, Figure 6; Figure 7C). At this point, the latch is returned to its resting location by the stored potential within the return spring; thereby fully enclosing the leaflets and completing the locking process (Figure 7D). To disengage the connector the latch is manually displaced by the user re-exposing the leaflets in a clockwise rotation disengagement process.

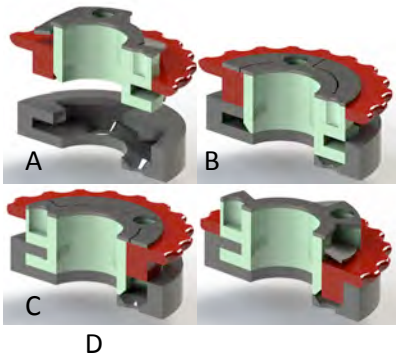

Figure 7: Assembly of Universal Coupler, Concept 2: (A) Unassembled, (B) Inserted but not engaged, (C) Engaged but not locked, (D) Full assembly with latch engaged.

Concept 3

Concept 3 (Figure 8) has an axial length of 10 mm and a 25 mm diameter. An important feature is room in the center for mounting two circuit boards, a distal board with male connectors and a proximal board for these to contact. The distal board mounts twenty-one gold plated pogo stick contacts, four for each of five conductor paths and a sixth single pogo stick in the center for the sixth path. Each pogo stick is rated to carry 2A. Thus, the five outer paths can carry 8A continuously. Typically, only two paths need to carry high current, and some of the pogo sticks can be omitted. During coupling of the two components, the sticks compress against gold plated bands on the other board, and then slide along the bands as the two elements of the coupling rotate 30° to their locking position.

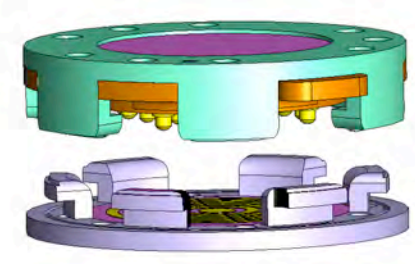

Figure 8: Universal Coupler, Concept 3. The complete concept 3 coupler is aligned for assembly. The pogo stick contacts are visible as are the five conducting rings on the proximal circuit board. The spring loaded lock ring is shown in the position it attains after the two halves are pushed together.

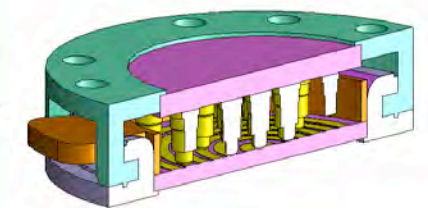

Figure 9: Assembled Universal Coupler, Concept 3. When the two halves in Figure 8 are pushed together and rotated 30°, the lock ring (orange) snaps into the position shown by the wing exposed on the left, preventing rotation. To uncouple, the user lifts (distal-direction) on the two locking ring wings and rotates 30° to the unlock orientation.

At present, the parts of concept 3 exist as a CAD model. This makes changes easy. For instance, there are six engagement elements on each coupling. If every element is 30° wide, there is a lock position every 60°. Perhaps users would prefer eight lock positions spaced at 45°. The circuit boards shown permit coupling at multiple rotational positions.

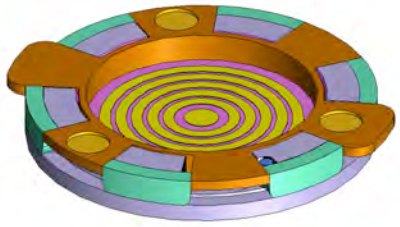

Figure 10: Section View When Locked, Concept 3. The tabs are not exactly 30°, which guarantees only one possible lock orientation. Making all angles 30° would allow six locking orientations. The three cavities shown are for springs that cause the ring to snap into the lock position just as the halves are rotated into alignment.

The universal coupler will seldom be used with devices that are only 25 mm in diameter. More typically, the coupler will join a size 7.75 hand to a fixed or a powered wrist. For cosmetic reasons, a two degree of freedom powered wrist should be oval where it connects to the hand and circular where it connects to the forearm, thereby maintaining a circular profile during axial rotation. Axial rotation would occur proximally between the wrist and forearm, while flexion-extension takes place in the wrist. This permits use of a coupler between the axial rotator and wrist with multiple locking positions.

An oval version of concept 3 is worth examining, because it illustrates the problems with scaling up the connections with the proximal and distal elements while always keeping all versions of the coupling interchangeable. As shown in Figure 11, there is a cavity between the two halves. Some of this cavity can be eliminated to make the lamination collar shorter.

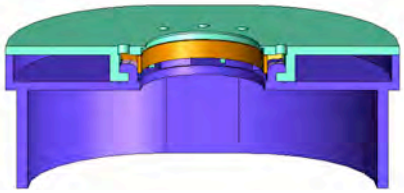

Figure 11: Oval Outer Profile, Locked, Concept 3. This section view show the lamination collar along with the distal oval element, which may be made integral with the hand chassis. The oval measures 44mm by 54mm and would be suitable for use with a size 7.75 hand.

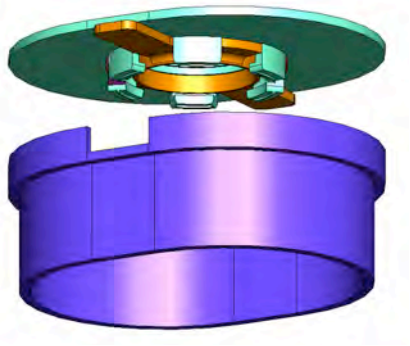

Figure 12: Oval Outer Profile, Unlocked, Concept 3. Note the need for slots to permit rotating the wings during lock and unlock.

CONCLUSION

There is a functional need for an improved universal coupler standard to meet the needs of patients using modern powered prosthetics. A standard adopted by the industry should allow the maximum function for the patient and the greatest compatibility between various prosthetic designs. The design requirements presented here outline what we believe would maximize function of a universal coupler. The three concepts show ways in which these design requirements may be met. The focus of these three concepts has been to show ways the mechanical aspects of a coupler design will meet the design criteria. Future work will include robust electrical connector features and minimizing how much moisture may get to the electrical connections. Furthermore, aspects of these three concepts may be combined to provide a single design with optimal function for the user.

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BREATHABLE LINER FOR TRANSRADIAL PROSTHESES

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ABSTRACT

The socket as the link between residual limb and prosthetic components is the crucial part of the prosthesis influencing the amputee's acceptance considerably.

In addition to protection from outside influences, the skin is responsible for regulating the body temperature. As parts of the skin are missing due to amputation, this functionality is minimized. This effect is increased by common liners consisting of silicone or equivalent materials, covering the residual limb surface.

A new developed liner is made of spacer fabric in combination with partial silicon coating for suspension. This way the functionality of the skin inside the socket is supported to regulate temperature based on permeability to gas and humidity. The cushioning effect of the liner reduces pressure peaks and shear forces to prevent skin breakdown.

The new approach of an interface design combines the comfort of using the conventional liner technique with the support provided by natural skin functionality.

INTRODUCTION

The skin of an adult person covers an area of approx. 1.6 - 1.8 m². [1] It is the largest human organ and a protective shield at the same time. The skin serves as respiratory, metabolic and protective organ. In addition, the skin supports the control of the body temperature. This is done in different ways. The release of heat by thermal conduction depends on the surrounding material. As humans prefer materials with low heat conductivity, loss of heat is low in this way. Heat release by convection and radiation is more effective and dependent on the difference of the temperature between body and environment. The dermis includes sweat glands. The sweat produced there is excreted by the pores of the epidermis located above. Due to evaporative cooling, the body temperature is effectively regulated independent of the ambient temperature. This requires that the sweat may evaporate which depends on the difference of the partial pressure of water vapour on the skin and in the air. Heat is transported away, even if the ambient temperature is higher than that of the skin which blocks other channels of heat release. As the limbs occupy more than half of the body surface, they also release more than half of the heat. [2] With the amputation of a limb, a large part of the body surface gets lost. The remaining part of the skin reacts with increased perspiration to balance the body temperature. [3, 4] In addition, the skin of the residual limb is covered by the prosthetic socket.

LINER

Liners have various tasks in prosthetics. They control disturbing forces and increase the wearing comfort of the whole prosthesis. In contrast to sockets without liner, the material properties of liners provide for enhanced suspension of the residual limb. In addition, donning is easier and more comfortable when a liner is used. Taking the plaster cast is facilitated; the handling for the patient is improved. Liners of silicone, polyurethane (PU) or copolymers (TPE) have become the fitting standard in the markets. Silicone has low elasticity and is breathable. PU absorbs humidity. Copolymers have high elasticity. These variants have in common that sweat cannot evaporate. In this way the liner inhibits the intended cooling process of the skin. In a survey conducted by van de Weg [5], patients report about the following three major problems that arise when wearing different liners. 26 % of the patients complain about perspiration when a liner is worn. 22.8 % have pains and 19.9 % mention problems with respect to unpleasant odour generation. It has to be pointed out that this study deals with prosthetic fitting of lower limbs. As the mechanical load situation is different there, the statements on pain development cannot be directly associated with prosthetic fitting of upper limbs. Perspiration and undesired odours, however, may develop with upper limbs too. Mak [6] describes that temperature and increased sweat production have negative effects for the patient.

Present liners are usually connected to the prosthetic socket by distal closure systems. This may lead to a "milking effect" as loading of the residual limb concentrates in the distal connection elongating the liner and the residual limb.

Another, non-distal connection between socket and liner could improve the wearing comfort. [7]

RESULTS

Textile

For a new liner concept, a special 3-dimensional textile spacer fabric has been developed. The side facing the skin is provided with bacteriostatic fibres that include silver ions (Ag+). The antimicrobial substance does not migrate into the environment. [8] The ions prevent the bacteria from multiplying [9] resulting in reduced development of unpleasant odours. The middle layer of the textile is provided with monofil threads forming a distance with damping function between bottom and cover layer. This effect is used for medical seat and bed padding to prevent pressure sores or for insoles [10, 11]. The monofil threads are provided with multifil fibres lying in-between. Due to their large surface, these Coolmax® fibres transport the moisture to the outside. The breathability of the textile is not impaired even in case of high humidity. On the side facing away from the skin the textile surface is provided with microfibres. Due to the large surface, these fibres have good capillary effects allowing for effective sweat evaporation.

New liner

The whole material includes a large air layer providing for low heat conductivity comparable to foam of approx. λ_{Sf} = 0.04 W/mK (table 1). Due to the temperature-isolating characteristics, only little heat is removed from the skin (convection is suppressed) so that the liner (figure 1) is perceived as pleasantly warm [10]. The low coefficient of static friction of the textile does not allow for the required suspension on the skin. This is compensated by partial silicone coating whereas the climatic effect is hardly limited. Produced sweat that may considerably reduce the static friction is transported away. The main objective is to create a functional combination of breathability and suspension of the liner on the residual limb.

Table 1: Examples for heat conductivity [12]

Material	Heat conductivity λ [W/mK]
Steel	45.0
Water	0.6
Silicone	0.2
Polyurethane	0.19
TPE	0.18
Spacer fabric, Foam	0.04
Air	0.0026

The circumferential elasticity of the liner is high. The liner is offered in different sizes to meet the individual needs of the patients. The longitudinal elasticity is very low. In case of tensile loads, shear stress on the skin is minimized. A distal pin has been consciously avoided. Force transmission is distributed to the whole liner. The space for the distal attachment mechanism is not required any longer. The liner may be simply cut to the needed length. Compared to common liners, the textile liner offers increased compressibility resulting in more physiological movements of residual limb and muscles. The optimized textile allows shortening without post-processing. Hygiene aspects have been realized in that way that the liner is easily washable.

Figure 1: Current functional model of the breathable liner

Patient trial

In a patient trial the individualization of the liner was conducted by simply cutting it to the right length. Due to the right elasticity of the material the liner provided a comfortable result for the patient in relation to compression and cushioning effect. In our case the patient had a distal bony residual limb (figure 2).

Socket

The arm liner becomes useable only in combination with an appropriately adapted prosthetic socket. The inner sockets of traditional transradial prostheses are made of deep-drawn thermoplastic, which provides both an intimate fit with the residual limb and sufficient strength necessary to support the terminal device. In practice, the shape of the socket is often difficult to control, resulting in a less-than-optimal fit. Moreover, a completely closed construction may lead to

donning, doffing and perspiration issues. A prosthetic socket with an open external frame has been developed to allow for easier donning and doffing, adjustment to variations in arm circumference. The socket along with the breathable liner could offer improved ventilation. An intimate fit is offered by a combination of residual limb-supporting flexible parts and the liner. The intention of the socket-design is to support the CPO for flexible and effective fitting of the residual limb.

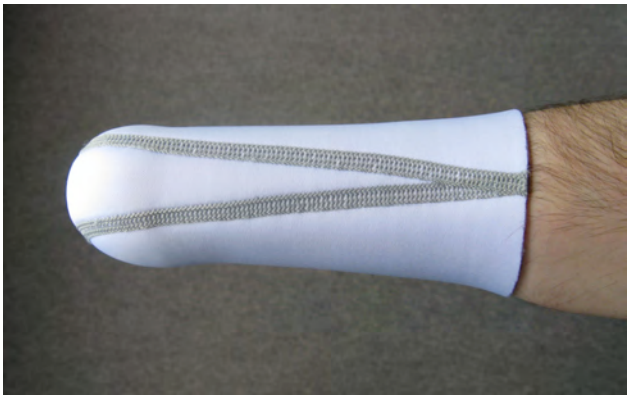

Figure 2: Individualized liner in patient trial

CONCLUSION

In contrast to existing systems, the new arm liner ensures the breathability of the skin. Humidity is transported to the outside to evaporate there. Undesired odours are reduced increasing the wearing comfort. So far the effects have been confirmed by a case study. The anatomic, large-surface attachment of the socket to the liner shall resolve the present problems of distal connections (milking effect, local loading, etc.). To confirm these effects in practice, further investigations including patient tests are required.

ACKNOWLEDGEMENTS

The authors gratefully acknowledge Anett Elsner for the preparation of this manuscript.

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BEBIONIC PROSTHETIC DESIGN

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INTRODUCTION

The *bebionic* hand by RSLSteeper is a fully articulating myo-electric hand with multiple pre-programmed grip positions. Increased function and hand flexibility provides design engineers with challenges in maintaining reliability and meeting increasing user demands. This paper will discuss the technical attributes, key design elements and the philosophy of their selection and evolution, as well as thoughts for the future. Special attention will be paid to thumb and digit design, feedback loop, material and hardware selection.

The hand is controlled in a similar way to other myo-electric hands by controlled muscle contraction. Electrodes measure electrical changes on the skin covering the control muscles, and instruct the five individual actuators within the hand to provide the desired movements.

The hand incorporates five high speed/force motors and is designed for low power consumption. The naturally compliant fingers and thumb of the bebionic hand provide a secure platform to perform everyday tasks using common grip patterns.

TECHNICAL INFORMATION

Principal Dimensions

Table 1: Principal Dimensions of *bebionic* Hands

	Principal Dimensions	Large Hand	Medium Hand
A	Middle Finger Tip to Hand Base	200mm	190mm
B	Thumb Tip to Hand Base	125mm	121mm
C	Max Chassis Width (no glove)	92mm	84mm
D	Diameter of Chassis at Wrist	50mm	50mm
	Palm Circumference (no glove)	220mm	204mm
	Maximum Opening width Tripod Grip	105mm with glove	105mm with glove
	Thumb Swing Through Angle	68°	68°
X	EQD only	5mm	5mm

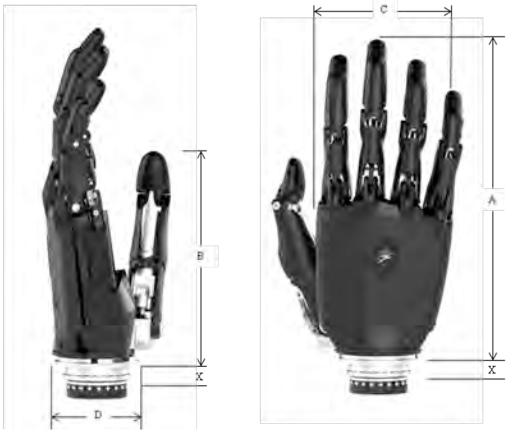

Figure 1: *bebionic* v2 Hand

Performance Specification

Table 2: Performance Specifications of bebionic Hand

	Large Hand	Medium Hand
Maximum Power Grip	16.8 ft lbs (75N)	16.8 ft lbs (75N)
Maximum Tripod Grip	7.6 ft lbs (34N)	7.6 ft lbs (34N)
Minimum Time to Close – Tripod Grip	0.4 Seconds	0.4 Seconds
Minimum Time to Close – Power Grip	0.9 Seconds	0.9 Seconds
Minimum Time to Close – Key Grip	0.9 Seconds	0.9 Seconds
Maximum Static Load – Hook Grip	70.5 lbs (32kg)	70.5 lbs (32kg)
Maximum Load Individual Finger – Hook Grip	35.2 lbs (16kg)	35.2 lbs (16kg)
Finger Tip Extension Load	13.2 lbs (6kg)	13.2 lbs (6kg)
Maximum safe vertical load taken through knuckles	198.4 lbs (90kg)	198.4 lbs (90kg)

KEY DESIGN FEATURES

The *bebionic* hand has many sophisticated design features providing flexibility for both user and practitioner.

Finger Design

The *bebionic* hand features individually powered, articulated digits each driven by and linked to its own 6V actuator. The actuators are positioned within the palm to provide a beneficial solution for weight distribution.

The actuators are assembled with individual separately programmed PCB's with onboard microprocessors. These are coded to constantly track the placement of each digit and provide precise control. This ensures accurate grip sequences and digits performing synchronized movements together every time.

An aluminum lead screw nut attaches the actuator to its individual finger proximal through carefully designed foldaway links. Made of nylon (Technyl A218V30) these foldaway finger links allow the fingers to flex freely and naturally, as well as allow the user to push up to 90kg through the hand to aid in standing from a seated position, a feature especially useful for the bilateral amputee.

In the event of mechanical overload it is important to protect the unit. The design accomplishes this with a shear pin fitted as part of the finger linkage. The shear pin is made from peek and is designed to fracture under a predetermined extension load, thus preventing damage to the finger and motor. The shear pin can be replaced locally by the clinical team.

The finger proximal moulds are shared between the index, middle and ring finger and are of an extremely robust construction utilising high impact thermoplastic, Hostaform C9021 GV3/20 (POM). All joint connections are locked with titanium fixed fasteners and square extension return springs. These materials maintain a high degree of strength and durability whilst reducing weight. Each finger also has tip mounted soft pads made from silicone. The palps are moulded in halves and shared between the index and ring finger.

Field replacements of individual digits can be done by carefully removing the shear pin, closing the hand slightly to lower the lead screw nut and then removing the hex screw from the knuckle base.

Thumb Design

The thumb also utilizes its own actuator and can be manually placed in one of two positions, opposed or non-opposed to the fingers. Located at the base of the thumb motor is a nylon (Derlin AF) trunnion nut which fits directly into the aluminum (AL-LM25TF) thumb bracket. A wishbone link, made from silicone brass, attaches the back of the bracket to the actuator to provide stability and allow movement towards and away from the palm.

The thumb bracket is attached to the chassis base through a fitted lower cam, mounting bolt and spring assembly. This robust assembly allows for continual manual movement of the thumb with a 68 degree range of baseline adjustability. The cam design allows the thumb to lock into an opposing or non-opposing position. This prevents the patient from wasting time aligning the thumb and allows for fast selection of grip patterns through the feedback loop (read switch). Also, when the thumb is locked it cannot back away under load, for example if the finger-object is hitting off center.

A read switch is located on the rear chassis near the outer side of the thumb bracket and works in conjunction with a small magnet located in the same vicinity on the bracket. When the thumb is in a non-opposing position the magnet is in close enough proximity to enable the read switch causing the non-opposed hand grips to be facilitated. With the thumb in opposed position, the magnet cannot enable the read switch and therefore the opposed set of grips are used.

Thumb alignment adjustment has been added to the design to allow clinical staff to reposition the thumb for additional grips or realign strike points as required by patient need. The thumb includes a medial-lateral adjuster to reposition contact with either the index and middle fingers or with the index finger alone. A slotted screw is fitted within the thumb pivot assembly and acts to adjust the friction on the internal clamp. To loosen requires turning two complete turns then pushing the screw inward. With the clamp slackened the thumb position can be manually re-positioned. The assembly must be retightened before electrically driving the thumb.

Another adjuster, located under the thumb base, is integrated to alter the baseline position of the thumb. This moves the thumb either towards the palm or away from the palm in order to optimize the contact point of the thumb tip against the opposing finger(s). Using a flat bladed screwdriver the adjuster should be turned clockwise to move the thumb towards the palm and counter-clockwise to move the thumb away from the palm. The effect will only be observed after resetting the hand (turning the battery switch off/on), not whilst making the adjustment.

Chassis Architecture

The rear chassis provides attachment of the thumb through the bracket, lower cam, mounting bolt and spring assembly. The selected *bebionic* wrist type is attached to the base of the rear chassis and held in place with three M3 x 8 SKT cap screws. The chassis is uniquely designed to conform to the shape of the palm PCB which is slid in and held in place with an M2 x 6 PAN POZI thread forming screw.

The front chassis design conformably houses the four finger actuators and a gear cover, while providing attachment

and placement for the knuckles. The gear cover (ABS material) protects the finger actuators from dust particles while also providing a shield for the palm PCB and wiring. Attachment of the front chassis to the rear chassis is done in three places. An M3 16 PAN POZI thread forming screw located in the outer palm of the front chassis attaches it directly to the rear chassis. An M3 x 16 SKT head cap screw below the first finger on the front chassis attaches it through the rear chassis thumb bolt. This also holds the thumb bolt/ bracket assembly in place. The third connection point is located on the back of the hand and includes a chassis link, T-bolt and M3 x 12 SKT head cap screw assembly that also contributes to further stabilizing the palm PCB.

The top chassis, or back cover, provides final enclosure and easy access to the finger actuators and palm PBC. An RF ID tag is placed on the inside of the cover to provide easy identification of each hand. At the top of the cover are three pins designed to slide into adjacent holes located on the front chassis beneath the back of the knuckles. The base of the cover is secured directly to the rear chassis with two M2 x 10mm trimet thread forming screw.

Program Switch

Located on the back of the hand is a membrane switch that utilises flexible PCB tracks to connect directly to the palm PCB. The tactile design allows the used to easily locate it beneath a glove. The switch has four functions and is integrated with selectable bleep and vibrate switch indicators, which can be activated through the *bebalance* software.

Grip Patterns

The *bebionic v2* hand can provide 14 selectable grip patterns / hand positions.

The thumb has two user selectable positions – opposed and non-opposed – with two grip patterns available in each position, thus providing four primary grip patterns. The user can sequence between the default and alternative grip pattern for each thumb position by applying an OPEN OPEN or co-contraction signal (depending on what setting is chosen within the *bebalance* software). To gain access to four further grip patterns, the user can alternate from the primary grip patterns, to the secondary grip patterns by pressing the program switch.

Several grip patterns – Hook Grip, Finger Adduction and Flat Hand – are achieved as partial grips of another grip pattern and therefore do not need to be actively selected. For instance, a partial close in Power Grip provides Hook Grip. A maximum of 10 out of the 14 grip patterns are therefore available to the user at any one time. Individual fingers can also be stalled manually by applying resistance, allowing further hand positions.

To achieve certain grip patterns it is necessary for the practitioner to adjust the thumb alignment so that the contact position between the finger(s) and thumb is changed. For instance, the thumb contacts on the index finger for Precision and Pinch Grip rather than the index and the middle finger for Tripod Grip.

This represents an increase of grip patterns compared with the version 1 *bebionic* hand, and demonstrates how new unique hand patterns such as MOUSE grip may easily be created.

Electronic Monitoring of Digit Position

The code sets a starting point for the counter when hand is first powered up. The number of revolutions of each actuator is counted in order to monitor placement of individual motors. This provides accurate and repeatable grip patterns.

Auto Grip Feature

Auto Grip is a selectable electronic feature that can be enabled or disabled through the *bebalance* software. It functions only with the thumb opposed and in Tripod or Pinch Grips. Once enabled auto Grip is activated by the user providing three consecutive close signals and de-activated when the hand is opened.

The rotation of the finger actuators is monitored every 50ms. Movement or slippage of a held object is detected by motor rotation. The appropriate motors are driven to prevent this movement occurring by changing finger position / grip force and therefore automatically providing a more secure grip.

BEBALANCE SOFTWARE

The hand is programmed using Bebalance software developed by RSLSteeper. Information is transmitted wirelessly to and from the system. A radio frequency transmitter / receiver module is incorporated within the hand. The software allows control parameters such as hand speed, grip force and grip selection to be individually optimized, set and stored. It also provides a range of control methods using one or two electrodes, or other inputs. The software provides real time analysis including adjustment of user signal and allows the user to practice using visual feedback. The software also allows a hand to be 'read' to determine its existing program settings.

FUTURE THOUGHTS

The *bebionic* wrist allows the patient to perform rotation, flexion and extension for either a left or right hand. It uses a single motor to accommodate both actions.

The wrist provides 230 degrees of rotation, 180 degrees of which is external (rotation to palm facing up) and 50 degrees internal rotation, and 30 degrees of flexion and 30 degrees of extension. Its diameter is approximately 50mm with a total length of approximately 75mm.. The *bebionic* wrist has an EQD connection, enabling compatibility with all *bebionic* devices as well as some competitor devices.

Figure 2: *bebionic* Wrist

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FIRST EXPERIENCES WITH THE VINCENT HAND

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SUMMARY

In this article a new prosthesis-technology will be presented. The System allows the design of single powered finger for partial hand prostheses and one of the world's smallest and lightest multifunctional hand system with 6 motors inside. Both applications are discussed with all possible features and benefits.

INTRODUCTION

In the past few years, a new trend has become recognizable in the field of hand prosthetics. The availability of smaller, high-performance drive systems, microcomputers, sensors and new materials is boosting the development of actively moveable, multi-jointed hand prostheses, often also described as "bionic" hand prostheses. Work on these systems is an inherent part of practical research in the meantime; three projects are described as good examples.

NIDRR – Powered Prosthetic Fingers 1989

This work was supported by the National Institute on Disability and Rehabilitation Research (NIDRR), and the Department of Veterans Affairs Rehabilitation R&D Funds administered through the VA Lakeside Medical Center, Chicago. The primary purpose of this project was to develop externally powered fingers [1]. Of particular importance is the work of R. Weir [2]. Weir's development shows for the first time the integration of a motor directly in a prosthetic finger. The feasibility of individually powered fingers developed with the introduction of motors only 10mm in diameter that are small enough to be placed within an artificial finger. The resulting design uses three motors, one each in the thumb, index finger and middle finger. In order to achieve the maximum pinch force, the thumb motor provides the speed, and the index and middle fingers deliver the force [3]. These fingers provide independent movement of different fingers and offer a solution for the limited available space in partial hand prostheses, especially for single finger treatments and all patients with a long hand stump.

KIT – Fluidhand 2007

The hydraulic working hand has been in development since 1999 at what is today's KIT [4]. Elastomer-based *flexible fluid actuators (FFA)* move in the different prototypes available between 5 to 8 single joints, whereby a soft, compliant grip is achieved. The system consists of a miniature pump, a bank of valves, a fluid tank, an electronic control system, a sensory force response feedback to the wearer of the prosthesis and a cordless PC-interface.

The FLUIDHAND is being further developed right up to the present day at KIT [5], see Fig. 1.

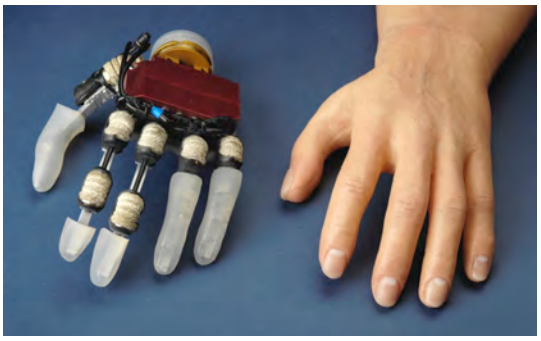

Figure 1: Fluidhand [S. Schulz, BioRobotLab, KIT, GER]

DARPA – RP 2009

The "prosthetic-arm project RP 2009" was started in the USA in 2005, under the leadership of the John Hopkins University and 30 other project partners. Today, the prototype of the prosthesis has 22 active joints driven by electrical motors, of which 15 alone are needed for the artificial hand, 7 for the shoulder, elbow and wrist.

NEW GENERATION OF POWERED MULTI – JOINED HANDS

Apart from the previously mentioned research and development projects, several manufacturers are about to develop multi-jointed bionic hand prostheses for the market, which allow separate movement of individual long fingers and thumb and their finger phalanges.

Touch Bionics – iLIMB Hand, ProDigits Finger

In 2007, the Scottish enterprise *Touch Bionics* was the first company in the market to launch the *iLIMB Hand*, an electrically driven multi-jointed hand prosthesis. A motor is located in each finger of the Hand, which directly

actuates the respective metacarpophalangeal joint. The metacarpophalangeal joint of the thumb allows manual swivelling to take place between the lateral and opposition positions. A second axis of the metacarpophalangeal joint is actuated via a motor integrated into the thumb and allows movement in the direction of grasp. Motor-actuated individual fingers are available under the name *ProDigits*.

RSLSteeper – BeBionic Hand

In 2010, the British manufacturer *RSLSteeper* introduced its latest prosthetic development at the 13th ICPO World Congress in Leipzig, the *BeBionic Hand*. Motors are located in contrast to the *iLIMB Hand* in the mid-hand (metacarpus). The fingers moves in the metacarpophalangeal and metacarpal joint. The thumb is brought manually into the desired position as for the *iLIMB Hand*, from which an active closing movement can be made. The control system of the prosthesis allows switching to be made between the individual types of grasp.

Vincent Systems – VINCENT Hand, VINCENT Finger

In 2010, Vincent Systems also presented a new prosthetic system at the 13th ICPO World Congress in Leipzig, which will be described in more detail in this article, based on [6].

VINCENT HAND GENERAL DESIGN

Components and functions

The Vincent Hand is a myoelectrically controlled hand prosthesis. It has the shape and size of a human hand and a particularly slender design of the fingers and the metacarpus, see Fig. 2. The very short structural height allows different hand sizes and stump lengths to be fitted, while maintaining anatomical proportions. Each of the four long fingers is equipped with its own drive. The metacarpophalangeal joint of the thumb is moved using two separate drives. The hand prosthesis has 10 actively moveable joints, which can be actively moved in the direction of bending and stretching.

Figure 2: Vincent Hand comparable in size to human hand

Altogether, the 6 motors of the hand allow active control of all essential basic types of grip, such as cylindrical, precision, hooking and lateral grip, index and key functions of the index finger, as well as a naturally acting normal position of the hand. Individual grasping movements e.g. for specific professional use, are also optionally available.

Material features

The precision parts of the prosthesis are made of high-tensile forged aluminium alloy, as used in aerospace applications. This provides the hand with an extremely high tensile strength and minimal total weight. Plastic and bronze bearings, as well as surface-coated steel axles and drives, ensure low-wear in operation and smooth running of the fingers.

Cosmetic – glove

Apart from the functionality and the weight, cosmetic aspects are one of the most important quality criteria of the hand prosthesis. The 10 actively movable joints, which achieve an extent of movement up to 90 degrees, respectively, necessitate a very high elasticity of the glove material.

The newly developed cosmetics consist of pigmented high temperature silicon with reinforcement in the finger pads and the inside of the hand. Particularly elastic joint areas, integrated fingernails made of silicon (in very high-quality models also made of acrylic), and the simulation of finger tip-like surface structures are some of the special functional characteristics. An inner glove made of silicon foam minimises the formation of folds and improves the adaptation characteristics, see Fig. 3.

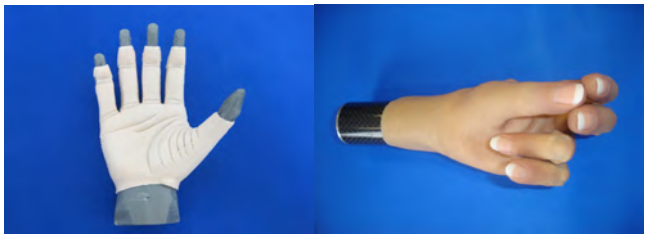

Figure 3: Inner glove with silicon foam (le.), silicon cosmetic glove (re.) [M. Schaefer, POHLIG, GER]

CONTROL OF THE PROSTHESIS

Control grip patterns

The prosthesis uses one or two sensors as standard in order to perform different hand movements. The user subconsciously controls each of the 6 articulation axes. Different grip patterns are already pre-programmed in the prosthesis to simplify its use – the user may switch between these. The user can also choose between different modes of switching, e.g., co-contraction, short single or double-signals to the "open" or "close" electrodes, as well as by a combination of long and short signals, fast or slow increase in signal intensity and combinations of all of these options. Opening and closing of the fingers takes place proportionally with each grip pattern. In the "learning phase" of the

prosthesis wearer, specific functions of the hand can be switched on and off.

Open interface

The protocol of the wireless interface is "open", i.e. this means that research facilities, orthopaedic specialist suppliers as well as prosthesis manufacturers can work with the Vincent Hand components. This function supports the global effort of the supply facilities according to an "*OPEN BUS"* standard, which supports the combination of prosthesis components from different manufacturers, so as to constitute an individual and optimal solution for prosthesis wearers.

User computer interface

A wireless interface of the hand to PC hardware, in connection with the service program Vincent Soft, allows the orthopaedic technician to adjust the controls to the individual requirements of prosthesis wearers.

Areas of use

The hand being presented is particularly well-suited for the fitting to small hand sizes, thanks to its slender design and its low weight. By use of elongated fingertips and the different sizes of inner and outer glove, scaling of larger variations of hands is made easier. The hand is a mixture of a cosmetic and a functional prosthesis, see Fig. 4.

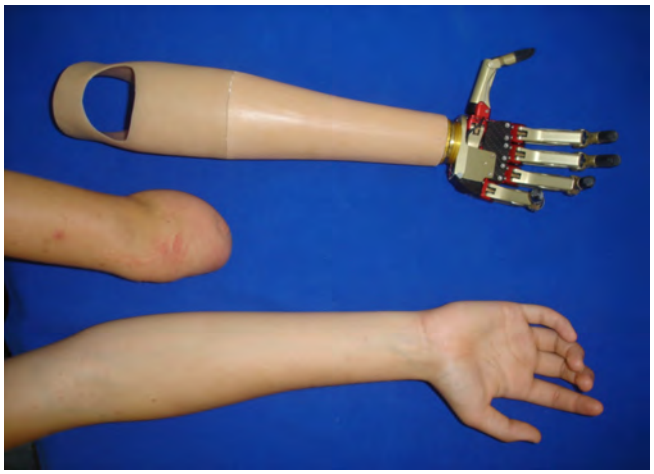

Figure 4: Vincent Hand and hand of a young female patient [M. Schaefer, POHLIG, GER]

The prosthesis has been designed for light tasks regarding grasping force and the ability to withstand component loads. Above all, manual worker prosthesis wearers who may exert very high mechanical forces on their prostheses will be better served with a conventional grasping tool. The hand is well-suited for most other tasks, such as e.g., in the service and office areas.

VINCENT FINGER FOR PARTIAL HAND

The provision of partial hand prostheses represents a particular challenge. The mostly very individual nature of the residual hand necessitates a prosthetic system that can be adapted to the different stump situations available. The restoration of a functioning hand is the main priority, although a second essential aspect is also the cosmetic appearance [7].

As both the number and position of the fingers to be replaced can vary in every patient, there is a need to be able to position prosthetic fingers individually. However, an elevated structural height of this individual finger in the basic joint would make the restoration of partial hands, in which the metacarpals are essentially still intact, more difficult. If the prosthetic structure is prolonged far above the natural anatomy of the hand, the thumb no longer reaches in opposition to the long fingers, which then also interferes with the functionality as well as the cosmetics.

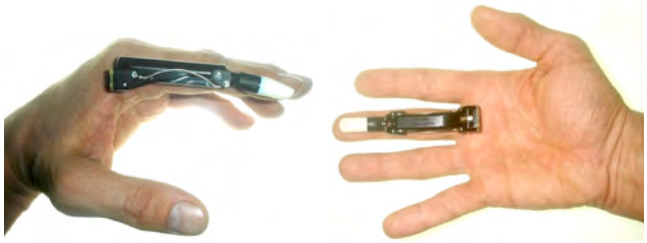

Figure 5: Vincent Finger comparable with a human hand

With the Vincent Finger, a very short-length single finger prosthesis is available, in which the structural height between the base plate of the finger and the first basic joint has been limited to a few millimetres, see Fig. 5. The final length can be adjusted via the exchangeable fingertips in 5mm graduations.

A 4-channel controller was developed for the control of the individual fingers, which allows the connection of two sensors, such as myoelectrodes or FSR-touch pads, as well as the connection and individual operation of 4 single fingers. As in the hand prosthesis, a PC interface is available which allows individual parameter adjustment.

Partial hand restorations

Single finger prostheses are currently in a phase of clinical evaluation with German and American partners. It was also possible to construct different partial hand prostheses.

Two patient cases will be presented here as examples. The first case describes a patient with a functioning thumb as well as part of the metacarpus, see Fig. 6. A prosthesis with four long fingers was constructed, whereby the short finger variation was used for the little finger. The control and battery system were placed in a silicon liner on the forearm.

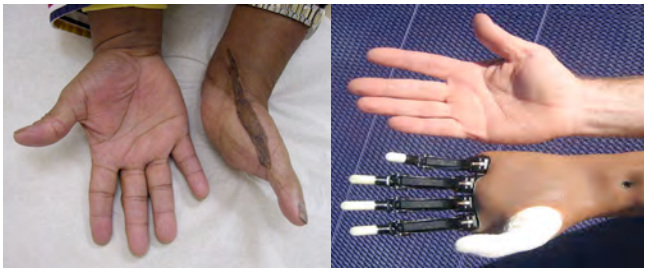

Figure 6: Patient left hand (li.), model of thumb and Vincent Finger (re.) [J. Uellendahl, HANGER, US]

The silicon cosmetics in the wrist area were designed to be very elastic; the thumb is able to move freely and can achieve an opposition position to the fingertips of the prosthesis long fingers, see Fig. 7.

Figure 7: Partial hand with Vincent Finger and cosmetic glove [J. Uellendahl, HANGER, US]

The second example depicts a patient who was fitted with a partial hand restoration, with a mechanical thumb as well as four long fingers, see Fig. 8. This can be brought manually into the lateral position or in opposition to the long fingers. The thumb is moved in this case on an arched track. The long fingers, on the other hand, are actuated via a myoelectric control system. Also, the integrated rechargeable batteries, charging point and control electronics are contained in a shaft placed on the forearm in this prosthesis.

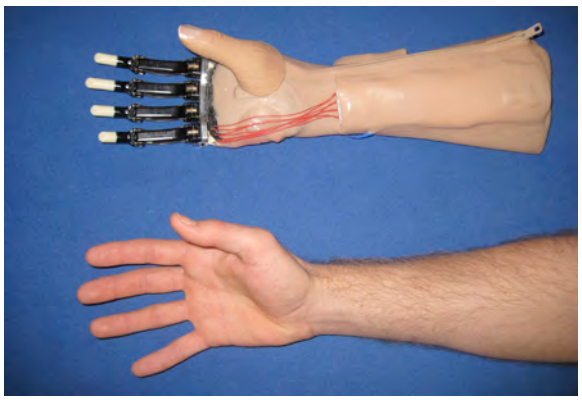

Figure 8: Partial hand with Vincent Finger and manual thumb cosmetic [M. Schaefer, POHLIG, GER]

A soft wrist transition permits full movement in this area. The manufacture of an optimally working, functional partial hand assumes considerable experience on the orthopaedic technician in this area [7][8]. The results of the first patient restorations show the great potential of the technology and encourage the further development of the system and its use.

ACKNOWLEDGEMENTS

Special thanks are due to Mr. M. Schaefer as well as the participating employees and patients of POHLIG for the numerous suggestions and feedback in the development of the systems which have been presented here. We would like to thank our American business partners, the orthopaedic technology association HANGER, especially J. Uellendahl, for support with the clinical trials.

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THE ELECTRIC TERMINAL DEVICE (ETD) – CASE STUDIES & EVOLUTION

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INTRODUCTION

The Electric Terminal Device (ETD) is a unique example of hybridization of body-powered with electric components, combining classic metal hook fingers with a high force motorized drive. Water-resistant housings allow use in wet/dirty environments.

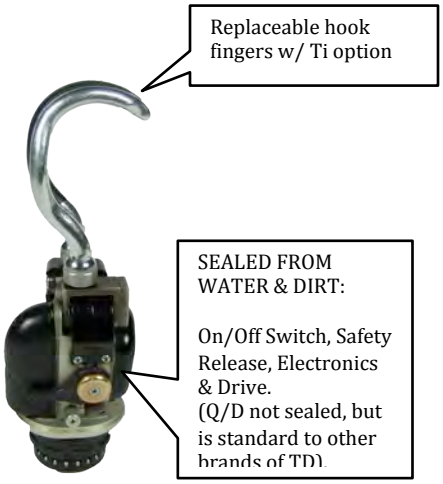


Fig. 1 – The basic ETD, introduced in 2002, is a hybrid of body-powered hook fingers (by Hosmer, Inc) with water-resistant housings enclosing a high-force motor drive.

RECENT INNOVATIONS IN THE ETD

Advanced ProHand

A new microcontroller, fully introduced in 2010, is mounted on-board and sealed within the water-resistant covers. The Adv. ProHand provides two functions which are unique within UE prosthetics, at present:

1. The *In-Hand (or In-ETD) MC Wrist Rotator* integrates the ETD and the wrist rotator, within a package only 2.9 cm longer than the ETD alone (The In-Hand version is 1.6 cm longer). The high-speed, high-torque rotator increases speed 2x over previously available wrist rotators.
2. An advanced *Brushless DC motor* (requiring eight contacts from the on-board microcontroller) can now be implemented into the ETD. The advanced motor increases the speed by 40%, while using standard battery and other electronics. Durability has been shown to increase greatly, since motor failure is greatly reduced in field trials to date.

The Advanced ProHand version of the ETD is fully compatible with virtually all other manufacturers' batteries and/or electronics, so it has in many cases supplemented the function of other TDs for work or hobby tasks.

Mechanical Flexion Devices

1. *Flexion/Extension Wrist* (passively operated) offers 50, 30, 0, -30 degree flexion/extension positions, with positive lock at each. Flexion allows gripping closer to midline and a very broad range of orientations.

Figure 2 - Flexion and extension allow the ETD to function as a positional vice, e.g., for fly-tying.

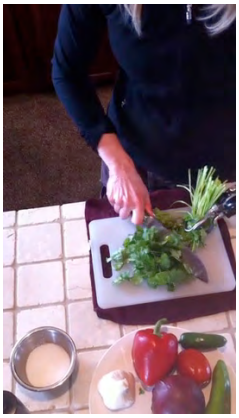
2. Multi-Flex Wrist (with spring-loaded return to neutral wrist position) allows the ETD to flex up to 30 degrees in all directions, returning automatically to neutral position. If needed, a mechanical lock also allows locking of the wrist in the flexion/extension direction at three positions: 30, 0, -30 degrees. A survey of Multi-Flex wrist wearers has shown that the lock is

important, but used for only 10-20% of tasks, since wearers prefer the comfort of the "natural-feeling" flexible wrist, and the security of grip it provides.

Figure 3 – While driving, the Multi-Flex Wrist reduces reaction forces on the socket, as the wearer pushes and pulls in various directions. Also, flexibility in the wrist allows a firm grip while the steering wheel moves and vibrates (as with this pickup truck).

Options for high strength:

Titanium hook fingers increase strength by ~200%, for heavy duty wearers. Ti fingers weigh 23 gm more than the standard aluminium, and are more costly, thus are optional.



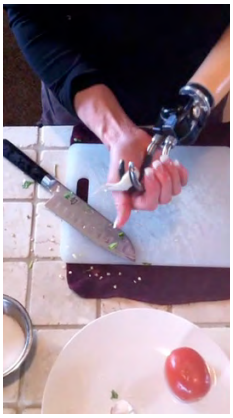


Figure 4&5 - Cooking tasks demonstrate the water-resistance of the ETD as well as the strength of the Titanium hook fingers.

Bilateral ETD wearers

Although in North America, the choice of electric prostheses is less common for bilateral limb loss, the ETD is often the exception to that rule. Adoption of hook-type TD shapes in the body-powered (b-p) arm, makes adoption of the ETD easier, so that changing from b-p to electric prostheses maintains very similar gripping shapes.

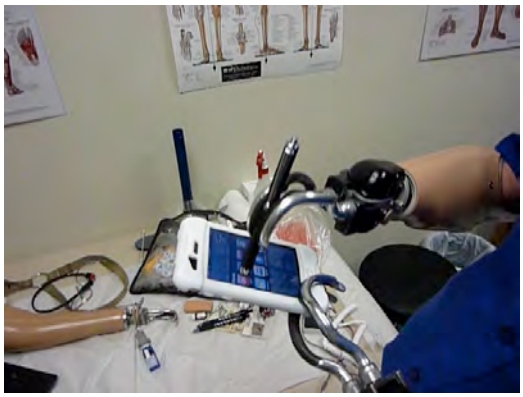


Figure 6 – Bilateral trans-radial wearer of ETDs, performs very independently – the touch screen is accommodated with a stylus.

RESULTS & CONCLUSIONS

A very wide variety of ETD wearers demonstrate a great range of functional usage. Innovations in drive mechanics, wrist positioning, and durability have increased functionality still more. Some conclusions are clear from our experience:

- Rugged work and hobby activities create a functional need beyond that available with hand-type TDs. Interchangeability of the ETD greatly increases the functional activities of electric prosthesis wearers.
- Stereotypes of hand vs. hook wearers are unreliable - male/female, rural/urban, blue-collar/white-collar , unilateral/bilateral characterizations do not predict the adoption or non-adoption of a hook-type prosthesis.
- Future design innovations will likely broaden the population using electric hook-type TDs, improving size constraints and strength, as well as aesthetic appeal.

ACKNOWLEDGEMENTS

Thanks to all the prosthetic providers who have fitted and maintained the ETDs in usage throughout the world. For the wearers pictured in this article the authors specifically thank: Ability Prosthetics (SLC, UT), and the Hanger P&O UEPP, with offices throughout the USA.

HYBRIDIZING BODY POWER & BATTERIES: DEVELOPMENT OF THE ELECTROMECHANICAL SURE-LOK CABLE CONTROL SYSTEM

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ABSTRACT

Within the field of upper-limb prosthetics, motive power for various components is typically provided by the user's own muscular exertion (body power, or BP) or drawn from external sources such as batteries. While prostheses "combining" these power sources are not unusual—myoelectric elbows with BP terminal devices (TD) for example—each system typically remains separate, operating independently of the other. An electromechanical version of the Sure-Lok cable control is in development to purposefully enable crossover or true hybridization. Melding the precision and versatility of electronic controls with the efficiency and ready availability of body power will make possible new prostheses offering maximized efficiency and functional utility. In operation, the new Electro-Lok uses existing, familiar electronic controls (e.g. myoelectric, bump switch, FSR, etc.) to govern cable-actuated BP components. Potential applications include electrically (un)locking cable-actuated elbows, modulating grip force in voluntary-opening (VO) TDs, sustaining grasp with voluntary-closing (VC) TDs, and improving control of multi-function wrist units, among others. New pediatric options are also envisioned. The Electro-Lok represents a new class of mechanisms intended to enable practitioners to hybridize and blend various prosthetic components in practical and intuitive ways for the benefit and enjoyment of their patients.

INTRODUCTION

Many amputees prefer the relatively lower cost, simplicity, and robustness of cable-operated prostheses for routine chores and engaging in recreational activities and hobbies [1-4]. Design inefficiencies in these cable-driven systems, however, often cause additional wear and tear of anatomical structures already compromised [4-6].

The pathomechanics underlying upper-limb amputees' overuse and repetitive stress injuries are not mysterious in origin. Viewed as an organic machine, anatomical structures of the body—such as the shoulder girdle—begin to wear out under repetitive, abusive loads. The supraspinatus, infraspinatus, teres minor, and subscapularis tendons of the rotator cuff, as well as the coracoacromial, trapezoid, and conoid ligaments of upper-limb amputees' contralateral sides are often found upon surgical exploration to be frayed, worn thin, or torn [4,6]. In addition to this mechanical damage, blood flow to muscle tissues—primarily the trapezius—is impeded by sustained harness pressure and inadequate muscle relaxation and recovery time. Muscle biopsies taken from amputees with chronic trapezius myalgia find coarsened fibers and changes in capillary vessel structure; under microscopic examination, investigators report these muscle fibers appear visibly "ragged" [7,8]. Finally, harness loop pressure on the axilla nerve and artery often causes axillary nerve dysfunction, impairing movement and sensation in the user's sound shoulder [9].

Unchecked, these effects are cumulative, incrementally degrading functionality and ultimately robbing amputees of their ability to effectively use their prostheses, with bilateral amputees particularly at risk [1]. In these cases, the outcome is a diminishing quality of life as users progress along a spectrum ranging from minor neck pain and stiffness towards debilitating, irreversible rotator cuff and shoulder girdle damage [7,8]. Myoelectric devices also load the shoulder, and are not a panacea [5,6].

Paradoxically, conventional prostheses' most desirable characteristics—intuitive and self-contained operation, extreme robustness, and preservation of physiological proprioception—are possible precisely because of the cable and harness [10]. Cable tension and harness pressure are inherent in the device's operation and cannot be easily eliminated. Their deleterious effects on the user's anatomy, however, can be partly mitigated by shielding users from having to sustain tension for prolonged periods.

As a first step towards this goal, the mechanical Sure-Lok was developed. Effectively a one-way cable lock, the device maintains cable tension while the user relaxes their muscles with an attendant reduction in harness pressure. Amputees using the device report markedly diminished fatigue and pain, and clinical practitioners are increasingly incorporating the Sure-Lok into new prostheses.

Core Technology

Figure 1 illustrates the underlying mechanism, called in the engineering vernacular a "unidirectional self-energizing friction brake."

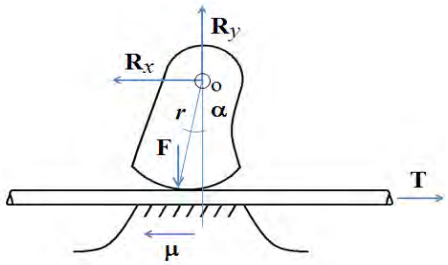

Figure 1: Mechanical schematic of a unidirectional self-energizing friction or braking cam

As shown, the cam rotates CCW about fixed pivot O to contact the cable. Given a coefficient of friction, μ, for the cam and cable interface, dimensional parameters r and α for the system can be chosen such that for any value of tension, T, the cam will generate a sufficient reaction force F to prevent cable movement in the direction of the tension—this mechanical arrangement is said to be "self-energizing" or "self-locking." Equation (1) describes this locking condition:

$$\mu \geq \tan(\alpha) \tag{1}$$

A significant benefit of this arrangement is that the cam provides cable locking action without the need for external power. Creating a practical embodiment, Figure 2, required detailed engineering analyses and laboratory experimentation, leading to several critical advances: 1) materials for the cam and structural base were found that generate sufficient holding friction without abrading or damaging standard prosthetic control cable; 2) a proprietary cam profile progressively compensates for both cable compression and material wear; and 3) special peripheral cam grooves distribute contact stress over the cable's surface, eliminating localized cable fiber chafing and fatigue. Commercial units have now

Figure 2: Mechanical Sure-Lok cable lock with top removed

been in service for over two years, and the patent-pending technology has been expanded into a next-generation design.

Reoriented Design

Originally designed as a retro-fit for existing prosthetic appliances that attached externally to the user's forearm shell similar to a watch, a new second-generation "vertical" design, Figure 3, was formulated that positions more of the mechanism "below deck" to achieve two objectives: 1) a sleeker and less obtrusive appearance, and 2) facilitating the addition of electromechanical actuation and remote actuation means (e.g. a chin paddle, lever, cable.)

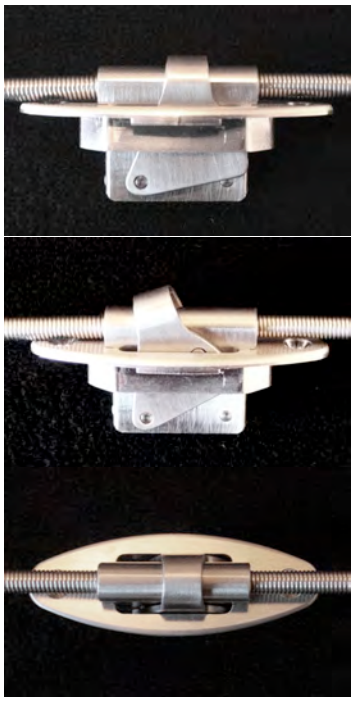

Figure 3: Virtical Sure-Lok disengaged (top), engaged (middle), and top view (bottom)

Testing of this new design is in progress with users reporting excellent intuitive operation and no mechanical failures. Given this configuration appears to be operating reliably, the development team has initiated work to incorporate electromechanical actuation.

Electromechanical Actuation

Envisioned is a complete cable lock system comprising the mechanical cam and its supporting structure, an actuator module that cycles the cam to engage or disengage, an electronics module that controls the actuator and interfaces with other commercial-off-the-shelf (COTS) sensors, and a small battery package if no other batteries are available to tap.

Two methods of actuating the cam are currently being explored. The first employs a micromotor servo unit, while the second uses electromagetic elements arranged into a bi-stable flip-flop configuration. Both designs, to ensure safety, let the user overide the actuation system to manually disengage the lock.

The design of myoelectric control circuits, locating electrode sites on amputees' residual limbs, and selecting muscle contraction/co-contraction actuation schemes for prosthetic devices are established practices within the O&P field. Circuitry to control the Electro-Lok will be designed to accept standard control signals. Moreover, prosthetists working with the development team are providing considerable clinical guidance to ensure the engineering design properly takes into account pragmatic clinical factors. The objective is to create a robust, modular commercial product that works well with existing standard components and that prosthetists can use intuitively with little difficulty.

Philosophy of Hybridization

Because energy is only consumed changing states (i.e. engaged to disengaged or vice versa), energy requirements for the cable lock are very low, and a small rechargeable battery can provide sufficient power to cycle the lock over an extended period. This a key aspect of hybridization: generally using external energy sources (that is, energy that comes at a premium) to control and direct the flow of the user's muscular energy (more abundant) in driving and positioning their prosthetic components. Each energy source is used more efficiently and effectively to benefit the amputee over a longer service period.

A device that mechanically locks users' control cables in response to myoelectric signals as contemplated here is just the first of several components being developed that are anticipated to open up the field of prosthetic hybridization. Others include control cable multiplexers that selectively "connect" one of several possible control cables to a user's harness, and a system for storing mechanical energy and returning it under electronic control.

Historically, both cable operated BP systems and externally powered devices have demonstrated their ability to improve the quality of life and well being of amputee users. It is reasonable to believe that new hybrid designs that capitalize upon and make available the best aspects of these two systems will substantially benefit amputees still more. The possibilities are unlimited and beguiling.

ACKNOWLEDGEMENTS

This work was funded by the United States National Institutes of Health (NIH) through a Small Business Innovation Research (SBIR) grant no. 1R43HD047039. The authors wish to thank Randall Alley, CPO of biodesigns, inc., along with Bill Hansen and T. Walley Williams of Liberating Technologies, Inc. (LTI) for their contributions to this effort. Opinions expressed in this paper are those of the authors and do not necessarily reflect opinions of the sponsors.

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CREATIVE SOLUTIONS TO BILATERAL UPPER EXTREMITY INVOLVEMENT

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INTRODUCTION

Individuals with acquired limb deficiency often experience difficulties with prosthetic fitting and use due to anatomical presentations that may affect both upper extremities. The non-amputated side may be affected by nerve damage, scar tissue or brachial plexus involvement. These problems may impact the potential user's ability to access power from available body movements. Signals from the nerves most commonly used to control utility of the prosthesis may also be impeded. These problems are likely to affect the capability of the consumer to comfortably wear and/or to access power or control from the prosthesis. Such problems directly result in challenges to use the affected non-amputated upper extremity, the prosthetic extremity and in difficulty to complete bilateral activities necessary for functional independence.

The Anchor technology was initially used to access scapular power on the same side of the trans-radial deficiency in order to operate the terminal device of a prosthesis. By accessing power ipsilaterally, lesser harnessing is utilized resulting in increased comfort, improved cosmesis and decreased axillary impingement. Other benefits include more symmetrical bilateral muscle development, decreased repetitive contralateral shoulder motion, and increased function particularly during bilateral upper extremity tasks. Consumers report the importance that intuitive movement rather than strategic motor planning is used to operate the prosthetic terminal device. For these reasons the prosthesis can then become a more natural extension of the body. This simple technology may also be applied to access power from other anatomical sources. These sites may include the forearm to activate a hand prosthesis or the trunk to stabilize a linear transducer. The cutaneous Anchor may prove to be a viable resource for suspension, stabilization or power of various prosthetic technology. This technology is in patent-pending status with the United States Patent Office. It has been used in patient treatment since August 2006. Patients appear to derive benefit and improved function of their prosthesis with the use of this device.

This paper addresses case solutions for problems associated with accessing power/control of the prosthesis using simple technology advances to complement the more complex technology used in the design of the prosthesis. Two case studies are discussed which include initial presentation with consumer-stated problems and concerns, solutions offered and training provided to the user from the perspective of the occupational therapist. Occupational therapists such as myself are concerned with the abilities of our clients to attain the skills necessary for maximal functional independence during the necessary tasks of life that include self-care, vocation and leisure time activities. Proficiency in these areas fosters enjoyable quality of life.

METHOD

Subjects

Two subjects are identified for the purpose of this reflection.

(A) is a male aged 45 years who experienced blunt trauma to his dominant right upper extremity at the trans-humeral level 4+ years ago involving the brachial plexus and resulting in amputation. He uses a trans-humeral prosthesis with a chest strap styled suspension, VariGrip hand and Otto Bock linear transducers and cables. A vertical strap to his belt is used to control the transducer which operates the hand. A transducer placed posteriorly connects to the chest strap and controls the elbow functions. (A) experienced loss of R sided vision as well from this trauma. He is a former semi-professional boxer and works as a fitness/boxing instructor at a gym. He was referred due to problems associated with accessing control of the prosthesis from available neuro-anatomy.

(B) Is a female aged 12 years from El Salvador who experienced electrical trauma to both upper extremities approximately 1 year ago. She sustained extensive burns and brachial plexus injury to her dominant right upper extremity as well as trans-humeral burn and amputation of her left upper extremity. She has residual scar tissue across most of

her volar and dorsal trunk as well as her right arm. Her right upper extremity has sustained decreased musculature as well as sensitivity. She presented for consideration of prosthetic technology. Although her left upper extremity appeared capable of enduring prosthetic componentry, her right upper extremity could not endure traditional harnessing at the axilla or at the chest. Higher technology of electronic or myo-electronic capacity was not considered due to the resources available in her natural environment. She attends school, plays soccer and enjoys reading and helping in the family home. She was fitted with a prosthesis featuring a TRS Lite-Touch hand and a mechanical elbow. She was referred due to the challenges of accessing body-powered control.

Apparatus

Variations of the cutaneous Anchor, utilizing technology for individuals with trans-humeral limb deficiency: (A) to stabilize linear transducers in volar and dorsal aspects of the trunk; and (B) to access ipsilateral body power for elbow and hand controls.

Procedures

Each client was fit with Anchor technology to suit individual needs: (A) 2 standard uni-button cutaneous Anchor pads positioned strategically to access scapular and lower serratus anterior control; (B) 1 double-button Anchor cutaneous Anchor pad positioned strategically to access scapular control. These controls activate both prosthetic elbow and prosthetic hand movement however through different mechanisms: body vs electronic. The prosthetist and the occupational therapist fit the client with the Anchor(s). Prosthetic training is provided which includes application, skin hygiene, use and care of the Anchor. Baseline observations are completed including clinical observations, video-graphed functional tasks. Follow-up videos of both clients are pending.

Data Analyses

Data is anecdotal via both therapist and client report including photograph and video-graph display. It appears to represent client satisfaction and ability toward maximal functional independence and ultimately toward positive quality of life.

RESULTS

Initial observations and results include active participation toward independence in self-application of both Anchor applications as well as prosthetic donning/doffing, success with prosthetic ability particularly as increased active spontaneous use, approved cosmesis and high client satisfaction. Consumer report reflects the intuitive nature of the movement required to effectively utilize the prostheses. This is important considering that both clients have experienced historical use of bilateral upper extremities and have also experienced absolute loss of one upper extremity and have experience impacted loss of some function in the other remaining upper extremity.

DISCUSSION

The cutaneous Anchor is simplistic in design. The parts are durable, easily available and affordable. The potential benefits of this technology appear to result in increased prosthetic wear and use (relative to frequency, tolerance, spontaneity) as it allows for improved comfort, cosmesis and ease of use during functional activity and particularly during bilateral activity The previously mentioned implications are proven to be beneficial for some individuals with trans-humeral deficiency, whether congenital (past proven) or newly acquired. It is clear that the technology is capable of activating both prosthetic hand and elbow; and is useable by populations of both limb deficiency and brachial plexus injury. Implications continue to project use toward dynamizing orthotics in clients with loss in function but not necessarily loss of limb. It is hoped that study of future work will prove this to be true.

ACKNOWLEDGEMENTS

Gratitude is extended to the clients who permitted the sharing of their stories, Shriners Hospitals for Children-Springfield, MA; Shriners Hospitals for Children International, HQ; and Michael Latour, Co-Owner, Single-Handed Solutions for continued support and encouragement. It is important to note the past beneficence and encouragement from Bob Radocy, TRS and Bill Hanson, Liberating Technologies, Inc.

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