MEC '08

Measuring Success in Upper Limb Prosthetics

University of New Brunswick's
MyoElectric Controls/Powered Prosthetics
Symposium

August 13th to 15th, 2008
Fredericton, New Brunswick, Canada

SYMPOSIUM PROCEEDINGS

Hosted by:
Institute of Biomedical Engineering

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Additional copies of the proceedings are available for $ 40.00 CD from:

Institute of Biomedical Engineering
University of New Brunswick
PO Box 4400
Fredericton, NB E3B 5A3
CANADA

ISBN: 978-1-55131-124-1

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Welcome to MEC '08

On behalf of the organizing committee and the staff of the Institute of Biomedical Engineering at the University of New Brunswick, we would like to welcome you to **MEC'08**. Once again, we are pleased to present a diverse and thought-provoking assortment of scientific papers and discussions relating to the field of upper limb prosthetics. Our theme for this year's symposium is ***"Measuring Success"***, an important topic for clinicians, researchers, and prosthetic manufacturers. The initial response from abstract submission and registration suggests that this MEC will exceed all previous symposia, and we hope you will find your time here enjoyable and stimulating.

The goal of the symposium is, as always, to share information, generate discussion, and inspire future research, which, in the end, will benefit all upper limb amputees.

This conference has evolved over the years to include pre-symposium courses and workshops to complement the three-day symposium. This year, we are hosting a course for qualified occupational therapists to learn to administer the *Assessment of Capacity for Myoelectric Control* (ACMC), which is a measurement tool that can be used in a clinic setting to measure function while using a prosthesis. There are workshops being offered by some of the leading manufacturers of prosthetic components, such as: Touch Bionics, Motion Control, and Liberating Technologies. There will also be a workshop on using motion analysis for the upper limb, as well as two workshops, one relating to Standardized Communication Interface for Prosthetics (SCIP), and the other regarding outcome measures in upper limb prosthetics.

This year's keynote speakers will highlight many of the advancements in research and technology pertaining to upper limb prosthetics. We are honoured to have as our Keynote Speakers ***Dr. Milomir Ninkovic***, from the University of Innsbruck, the Clinical Team from the ***Rehabilitation Institute of Chicago***, (specifically, Robert Lipschutz, Blair Lock, Laura Miller and Kathy Stubblefield), and ***Mr. Stuart Harshbarger*** from the Johns Hopkins University Applied Physics Laboratory.

We hope you will join us for the conference social events on Wednesday and Thursday, August 13th and 14th. Social events are an important part of MEC, as they allow time for informal networking and discussion of the day's events, while experiencing some of Fredericton's warm hospitality.

MEC benefits from the financial support of a number of sponsors, and we thank the leading manufacturers, provincial, regional and federal organizations for their continued support.

Once again, welcome to **MEC '08**. MEC is a team effort by all of the staff and students of the IBME, so please don't hesitate to ask for assistance of any of the team members.

Wendy Hill **Peter Kyberd**

Co-Chairs MEC '08

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The Institute of Biomedical Engineering and the *MEC' 08* Organizing Committee would like to recognize the following organizations for their contributions to the symposium:

CIHR (Canadian Institutes of Health Research)

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Touch Bionics

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Thank you for making this symposium a success.

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Principles of Rehabilitation after Bilateral Hand Allotransplantation

Marina Ninkovic

Abstract

Background: The previous experimental results with long-term survival of animal limb allograft and positive results of single hand transplantation encouraged us to plan (January 1999) and perform first (March 2000) and a second (May 2006) bilateral hand transplantation as well as first (February 2003) bilateral forearm transplantation.

Method: Comprehensive evaluations of all recipients(first male patient lost both hands in carpal level after a blast explosion accident, second male patient with bilateral proximal forearm amputation after a high-voltage burn accident and the third patient with bilateral amputation at the distal forearm after a blast explosion accident) and search for adequate donor according to the sex, size, skin colour and texture were done preoperatively. Three surgical teams worked simultaneously and independently. The immunosuppressive therapy was started intraoperative. At the first day after successful surgery the intensive programme of rehabilitation was introduced. This protocol was designed based on early protective motion program and on specific cognitive exercise program after Perfetti. Specific tests to assess sensory and motor recovery were performed each month for the first 12 months and then every 3 months thereafter.

Results: There were no intraoperative or early postoperative complications. Sensory and motor recovery increased gradually and cortical reorganisation were shown. Clinical examination and electromyography evaluated recovery of the intrinsic muscles of the hand. Judging from the Tinel sign and Semmes-Weinstein Monofilamententest advancement, nerve recovery appeared to be ahead of schedule as compared with comparable replant patients. Ten months after surgery the first patient was able to return to his job and he is complete independent in the activities of daily living., as well as our second patient. The third patient is still in rehabilitation’s program showing adequate motor and sensitivity recovery concerning the level of the amputation.

Conclusion: Limited experience of hand transplantation confirmed importance of adequate patient selection, recipient donor matching, minimal cold ischemia time, as well as appropriate simultaneous surgery, immunosuppressive therapy and early intensive and long-lasting rehabilitation. These prerequisites can lead to effective and save hand transplantation with very promising functional and andaesthetic results.

Targeted Muscle Reinnervation and Occupational Therapy in various rehabilitation settings

Shawn Swanson, OTR/L
Otto Bock HealthCare

Targeted Muscle Reinnervation (TMR) is a new surgical technique for transhumeral or higher levels of upper extremity limb loss. The rehabilitation team at RIC has seen great success with their patients. RIC has a unique opportunity in their research environment as it allows for close supervision with socket modifications, therapy training and component/software program management (prosthetists, engineers, and therapists on site and patient attends training for several hours a day for an extended time period).

In the United States, TMR is also being done outside the research environment in the following locations: Harborview Medical Center in Seattle, WA, Walter Reed Army Medical Center (WRAMC) in Washington, DC and Brooke Army Medical Center (BAMC) in San Antonio, TX.

This paper will discuss and share experiences outside of the research setting in the US. The two other settings are military and healthcare/insurance.

The military setting will allow for close supervision as well (prosthetists and therapists on site and have treatment/training once a day), but not as close as the research environment. The soldiers take leave from the hospital, have various family/military obligations and could possibly have other medical issues that interfere with treatment/training progression.

The healthcare/insurance setting does not allow for close supervision as rehabilitation will typically be on an outpatient basis. (Prosthetists and therapists aren't necessarily in the same building and treatment may be for 1 hour/day times 2-3 times a week secondary to insurance limitations).

The goal in the near future is to bridge the current gap between research and "insurance world" applications.

The Role of Mental Health in O&P

Ruth M. Morris, LMSW

Why should O&P practitioners be concerned with the mental health of patients? A thorough understanding of issues facing amputees will assist prosthetists, technicians, and other professionals involved in rehabilitative care in providing comprehensive care, which will ultimately lead to successful prosthetic use. Practitioners will be given an understanding of the potential liability risks that may arise with patients in need of mental health attention, in addition to basic skills to identify when to refer patients to a mental health professional. An overview of the psychological impact of amputation and the various ways issues may manifest in the rehabilitative process will enlighten all care professionals involved to provide appropriate treatment and collaboration among disciplines.

CASE STUDY: REGAINING INDEPENDENCE, A PATIENT WITH BILATERAL UPPER EXTREMITY LIMB LOSS

Kristin Gulick, OTR/L, CHT
Director of Therapy Services
Advanced Arm Dynamics

As advances in prosthetics are accelerating in the laboratory, research, and clinical settings, it is critical for the developers and medical providers to understand the setting in which the client with upper limb loss actually lives and functions. The perspective that we gain from observing the client in their unique environments should drive all that happens in our research and development labs and patient care clinics. A client's ability to function when assessed outside of their environment is falsely impacted by challenges that they may not have to deal with in their environment or there may be modifications in our clinics to ease function that are not available to them at home.

This case study presents a client with bilateral upper extremity limb loss. The focus will be on the rehabilitation that occurred in his home. The client is a 43 year old male who moved to the United States from Mexico 23 years ago. While Spanish is his primary language he does speak functional English. He is married and has four daughters under the age of 13. He has worked for Anderson Hay for 22 years. His job there was to operate a machine that takes normal sized hay bales and compresses them to 1/3 of normal size for shipping overseas. On June 2, 2006 he lost both upper extremities, the right at the transradial level and the left at the transhumeral level in this machine. He was hospitalized for three weeks and discharged home with home health nursing for wound care, occupational therapy, and a home health aide. Within one month he began to attend outpatient occupational therapy for UE rehabilitation and pre-prosthetic training. He was first fit with a preparatory myoelectric prosthesis on the right side on 9-28-06. He began basic prosthetic training with an electric hand and wrist rotator. He switched modes using 4 channel processing and was able to successfully do this immediately with fitting due to pre-prosthetic training with the MyoBoy system and his inherent motor abilities. He required wrist flexion to perform ADL's at midline, however he did not want to use a non-anthropomorphic terminal device, and therefore an electric hand with wrist flexion was provided. Self-feeding was the first ADL that the patient wanted to focus on and he quickly achieved independence with fork foods. Meanwhile, care in the home was not going well due to trust and cultural issues between the client and his family and the care providers. At that point this therapist began to provide therapy in the home.

The initial home visit opened our eyes to the true challenges for this patient. The patient had just purchased his first home in the United States and was very proud of the home. He wanted to provide his family with a residence that was "theirs" and some land to have a garden and some chickens. Due to financial limitations, they purchased a "fixer-upper" and the patient's relatives were going to help with restoration in their free time. The house did not

have running water or a refrigerator for 2 months. The main room floor and sub floor was removed for a number of weeks, requiring one to walk across 2”x 4”’s while viewing the crawl space below. There was a 6 week delay in getting the foot operated bidet installed in the one bathroom that was shared by 6 people. The challenges were too many to list and required a very individualized, goal oriented approach with success measured in small increments. The goals were based on the patient’s desires regarding self-feeding and toileting at the top of the list, followed by home access and garden activities, and bathing and dressing subsequently. Everyday items, adaptive equipment, and custom modifications were made to enable this patient to be independent in his ADL’s and IADL’s. The process is ongoing as he expands his abilities, interests, and requirements.

This patient’s rehabilitation was significantly impacted by his home environment and by his cultural belief system. It is critical to the success of our patients that we fully understand their physical, emotional, and cultural world and base our interventions on this understanding. The opportunity to provide care for this patient in his home resulted observation of his interactions with his family, friends, and community. This knowledge impacted his care by providing us with information about what type of terminal device he would use based on cultural values and assisted us in prioritizing goals. Often times the medical provider’s bias about the importance of certain goals can impact the actual success of the patient and the perceived success of rehabilitation by the team. For this patient, independent bathing and dressing were not as important as some IADL’s. The patient’s decreased motivation for independence in bathing and dressing was concerning to the team, however this is the reality of his life and culture and does not decrease success of his rehabilitation.

In this current market of third party payers basing payment on achievement of goals and independence in activities valued by American society, there is concern for the person who has a different value system or lifestyle. By describing some of these unique situations we may assist the rehabilitation process by increasing understanding of the client’s individual goals. Working with a client in his or her personal environment opens our eyes to the true ways that he or she uses prosthetic devices. This experience is invaluable for the medical provider and for the client. Communication from those of us who provide treatment in the client’s environments to researchers and manufacturers will result in the continuum of improvement in care of the client with upper limb loss.

"The Knowledgeable Prosthetic Limb User: Treatment Techniques for Occupational Therapists."

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The most important skill an occupational therapist can give an adult upper extremity prosthetic user is the confidence that the prosthetic limb can help them efficiently perform any task they encounter. Ultimately, the client will decide how they wish to use his or her new prosthetic limb, but for a brief time the occupational therapist plays an important role in educating and coaching the client. It is the therapist's role to expose clients to the full range of methods that they can use their new arm for functional tasks and to facilitate good problem solving skills. Having limited treatment time, it is crucial that treatment is efficient and focused on what will benefit the client most. When someone is learning a new skill, he or she tends to seek as much assistance as available and transitioning someone to independent analysis and refinement of performance is challenging. Clients who demonstrate good problem solving skills and understand task demands are more inclined to translate these clinical skills into functional tasks and fully integrate the prosthetic limb into their daily lives.

People who use prosthetic limbs have unique desires for the use of their devices. Some clients state cosmetics as a primary concern and may tend to keep it out sight as much as possible, while some want to perform as many tasks as possible with their prosthesis. Therapy needs to be client based and the use of the prosthetic limb needs to fulfill what they desire the limb to help them accomplish. It is very important that all patients be exposed to advanced prosthetic limb skills early in therapy because they may limit their potential use of device without this exposure. The new prosthetic limb users need to have knowledge of advanced skills of the prosthetic limb, as their demands may change in the future. The prosthetic limb user will need to adapt to new tasks with the prosthetic device throughout his or her life. As upper limb prosthetics advance, it is instrumental that the knowledge of the client advances as well. The occupational therapist needs to assist these clients in developing comprehensive knowledge of the device and independent problem solving skills as devices change and clients' needs change. Balancing user demands and knowledge of prosthetic limbs is vital to successful integration of the prosthetic limb throughout the client's lifetime as needs and activities change.

Education is essential early in pre-prosthetic training. The client needs to have a realistic understanding of what his or her prosthetic limb will be able to perform and how it works. Myoboy from Otto Bock is an excellent tool to build the muscle coordination skills of the client initially. It also provides many opportunities to educate the client and integrate him or her into understanding the operation of the prosthetic limb. A therapist can have the client palpate for contractions and try various types of contractions. This also facilitates the client reintegrating the amputated and painful limb into his or her body scheme. Wound care, desensitization, strengthening of proximal joints and scar massage also assist the client in becoming comfortable with the newly altered limb. The client should be encouraged have knowledge of where his or her best myosites are located, and what level of amplification is working best for them. This is an excellent time to educate the client about the components of the prosthesis. The therapist needs to be mindful of the client's threshold for education as they may not be ready emotionally after sustaining this type of traumatic injury. The therapist does not want a client to interpret prosthetic training as an emotionally difficult experience. Training needs to be balanced given mental availability of the client while demonstrating the advantages of use of a prosthetic limb. Given more knowledge of the prosthetic limb allows the client to become more comfortable with the device and orients them into problem solving role.

Initial training with an upper body prosthetic limb should provide the right amount of physical challenge and problem solving challenge. Any time a person is going through learning new coordinated movement, he or she tend to seek out support and numerous cues to perform new tasks appropriately. For example, one does not usually learn a to salsa dance in one day and he or she tends to seek out ways to have more support before they dance before a crowd. Some clients may need continuous step by step instructions, but the real challenge for the therapist is to find ways that clients can safely problem solve on their own without becoming overwhelmed. This facilitation of independent problem solving assists the client in achieving some comfort with experimentation and gain confidence in the use of his or her prosthetic limb. Repetitive simple grasp and release tasks or rote exercises are great ways to assist a client in gaining some confidence with the device. This is the time to begin addressing the components of good prosthetic limb use, including prepositioning the terminal device and incorporating normal quality of motion and body mechanics. Occasional challenges can help them begin to take ownership of the function of the prosthetic. For example, asking the client to find all of the ways the prosthetic limb can be used to assist him or her in opening a

plastic bottle. Quizzing a client is a good way to facilitate problem solving, ensure retention and assist the client in active use of the vocabulary of prosthetics. As a client's skill progresses the therapist should attempt to limit cues to facilitate independent performance and problem solving of tasks outside of the clinic.

As a client becomes more confident and comfortable with the use of his or her prosthetic limb, therapy can become more challenging. Stimulating analysis of tasks is important in this stage. Stimulating the client's need to anticipate positional changes while completing a task with finesse is also important here. A good question to ask is, how many different methods can you find to use the prosthetic limb in this task, and discuss what method works the best for them. For example, asking the client to find as many ways as possible of holding a fork with a sensor hand while cutting. The goal of this is to build the client's repertoire of skills with the prosthetic limb, it may not be the most effective way at present but that skill may be utilized in a different task in the future. Unilateral performance with the prosthetic limb is a pathway to this creative independent use, although maybe more time consuming, it improves the client's skill and dexterity with the prosthetic limb. For example, assembling legos without use of the unaffected side is a way to increase light touch and gentle manipulation. Other questions include; what terminal device is best suited for this task? Would a different type of object be better or worse, such as opening a glass or plastic bottle? Activities that are relevant and meaningful to clients are valuable therapeutic tools in this stage as clients find new ways to integrate the prosthetic limb into meaningful tasks of their life. Talking through the steps of a novel task is possible here too, such as describing the steps he or she would perform to start and operate a weed wacker. Even if the item is not available in the clinic it fosters the anticipation and fluid problem solving necessary for successful and timely integration of the prosthetic limb into an activity. Clients need to be familiar with minor adjustments and repairs of their prosthetic limb and how to adapt and overcome when a device prevents them from completing a task or when the device inevitably fails. There are many ways to stimulate patients to achieve more problem solving abilities and skills with their prosthetic device.

The client will also benefit from education in adapting the task or terminal device to accomplish certain tasks. Occupational therapists have a comprehensive knowledge of this and there are a number of common items that the patient could benefit from knowing how to use. Attaching loops of fabric to items can assist in performance with terminal devices. Having velcro strapping available can

improve the control of objects. Non-slip material can assist the patient in a variety of tasks. Creating a groove by building up material on an object can improve grip control of a slippery object. Splinting material is excellent for adapting the terminal device or the object. For example, splinting material can be molded to wrap around a joystick on an Xbox controller to extend it or create a surface more compatible with a terminal device. Adapting individual specialty sporting terminal devices is also important to ensure best performance. The more adaptable clients can be with tasks, the more likely they will achieve success in physical tasks that may improve their quality of life and comfort level with dealing with physical challenges.

People who suffer from limb loss will have many of challenges both physical and emotional throughout their lifetime, and they rely on prosthetists and occupational therapists to set them up for success with the use of their prosthetic limb. These clients need to be empowered to find what works for them the best and assisted in developing advanced skills with their prosthetics. Clients with a comprehensive knowledge and improved ability to problem solve activities will be prepared to take on new challenges with their prosthetic limbs. It is the role of the occupational therapist to help these clients find these pathways to success and ease their return to a productive life. Clients who demonstrate comprehensive problem solving skills increase their chances of comfortably integrating their prosthetic limb into daily life.

UTILIZATION OF NEGATIVE PRESSURE FOR SOCKET SUSPENSION IN UPPER EXTREMITY PROSTHETICS

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ABSTRACT

Negative pressure, or vacuum, has been utilized for suspension in lower extremity prosthetics with impressive results. The authors describe a technique to enable the application of vacuum in the suspension of a transradial prosthesis. A discussion of the merits and challenges of such a system follows with emphasis on troubleshooting and common causes of failure. Examples of this technique in various applications will be presented. Future innovations and potential for further development of this technique for other levels of upper extremity amputation will also be offered.

INTRODUCTION

The negative impact of poor suspension is a primary concern for many prosthesis users[1]. Improving suspension can lead to enhanced function, increased comfort, decreased perceived weight of the prosthesis, and potentially decreased skin trauma. The advantages of elevated vacuum socket design for suspension are well documented in the area of lower extremity prosthetics.[2,3,4] Numerous systems are currently commercially available. Our clinical group is exploring these concepts for application to upper extremity socket design at all amputation levels up to and including transhumeral.

This paper presents a technique for elevated socket fabrication for the transradial level. Eight patients fit with the elevated vacuum socket design for a task specific application. These eight transradial fittings serve as a “proof of concept” that vacuum sockets are effective in suspension of prostheses typically requiring considerable resistance to forces of distraction. While these early fittings have not yet incorporated electric components or control strategies of electric prostheses it is clearly a next step in the developmental process that requires attention.

SELF-SUSPENDING TRANSRADIAL SOCKET DESIGNS

Many transradial self-suspending sockets designs have been developed and used in an effort to eliminate the need for a harness for suspension. Variations of self-suspending designs are widely used in upper extremity prosthetics today, especially in myoelectric devices. Examples of self suspension utilizing the elbow bony anatomy are the Muenster technique [5], the Northwestern University socket[6], and the Trans Radial Anatomically Contoured (TRAC) socket[7]. Other socket designs have utilized elastic materials such as silicone to create suction between the socket and residuum. Examples of this type of suspension are roll-on liners[8,9] with locking mechanisms or lanyards, and custom silicone sockets.[10]

Elevated vacuum systems for suspension are an evolution of suction systems. The larger the air volume remaining in the sealed socket the larger potential for motion is present. By removing this air and creating a vacuum internally, motion between socket and residuum is virtually eliminated[3].

PATIENT SELECTION CRITERIA AND TYPICAL CURRENT APPLICATIONS

Eight patients were fit with an elevated vacuum socket for use with an transradial activity specific prosthesis. Each of the patients was previously successfully using a self suspending socket design for a myoelectric prosthesis and a roll on liner with pin lock with auxiliary suspension harness for a body-powered prosthesis. Each patient requested a task specific prosthesis that would reduce pressures on the bony anatomy of the elbow or eliminate the confining harness during specific activities. All agreed to be fit with an elevated vacuum system.

The general indications for fitting were: a well healed mature residuum that had already successfully tolerated the suspension of both a self-suspending socket design and a roll on sleeve with pin-lock design. The length and shape of the residual limbs were not used as selection criteria. Caution was exercised for limbs with deep invaginations as the effect on the skin folds under vacuum were unclear.

FABRICATION AND FITTING

Casting and Model Rectification

The patient is fit with a silicone roll-on liner with no internal matrix or distal pin. Both custom and off the shelf liners were used in our cases depending on the uniformity of the residuum shape. The off the shelf liner used was the ESP Streamline. The residuum is positioned with the muscle tissue relaxed, the elbow joint in slight flexion, but at rest, and the patient standing with the shoulder joint relaxed. A container of alginate is prepared that is tall enough to cover the residuum to a point two inches proximal to the elbow. The limb is positioned in the alginate and the alginate is allowed to set with the limb relaxed. The resulting negative mold is filled with plaster. The model is gently smoothed with no build-ups or reductions in the rectification process. There is no attempt to create a volume change of any kind during this rectification. It is our intention with this technique to have the tissue of the residuum come into full and firm contact with the hard socket prior to the introduction of vacuum. It is important that when vacuum is applied, that the soft tissue is not distended, distracted or otherwise stressed in the socket.

Test Socket Fabrication and Fitting

A test socket is blister formed from a clear plastic such as DurrPlex. The socket should be thick enough to ensure rigidity at the proximal brim. The socket is removed from the model and the trimlines established. The trimlines for the finished socket are distal to the olecranon process, the medial and lateral epicondyles, and the cubital fold. A vacuum tubing barb (Otto Bock 4Y300) is installed into the distal aspect of the plastic by drilling and tapping the plastic. The threads are sealed with a quickset two part epoxy during installation. A one way auto expulsion valve(Otto Bock 4Y311) is then installed using vacuum tubing to connect it with the barb on the socket.

The patient dons the same silicone liner used during casting. An interface to wick out trapped air such as a nylon sock must be used between the liner and the plastic test socket. This also allows the residuum to slide into the test socket without excessive friction. If a custom liner is being used, this interface can be integrated into the outer surface of the liner. The most important aspect of this test is to ensure total contact of the liner and residuum to the test socket without the introduction of vacuum. This point cannot be stressed enough and the success of the system depends upon this total contact. Total contact must be achieved with a minimum of force. No gapping proximally or voids distally can be present. After this is verified, a sealing

sleeve (Streifeneder 3s10/s) is applied higher than most proximal portion of the nylon wick. A hand vacuum pump (Actron CP7830) is used to produce the internal vacuum. Vacuum of up to -25inLb can be applied. The comfort of the socket can now be evaluated by pulling distally, pushing proximally, medially, and laterally. Rotational stability should also be confirmed.

Fabrication

1. Obtain a positive model by pouring the test socket with plaster and blister form an 1/8" layer of DurrPlex over the model
2. Abrade the surface of the plastic prior to an initial carbon lamination. This lamination is meant to be thin and light. One layer of braded carbon is sufficient
3. Created the outer frame shape out of plaster. Include a recess on the medial or anterior aspect deep enough to set the valve barb below the surface of the frame. Install a lamination dummy the appropriate size for the wrist to be installed later
4. Apply a PVA bag over the shaped plaster and inner socket. Laminate an outer frame using a preferred material lay-up. 3 layers braded carbon and nylon separators were used for our cases
5. Trim frame proximal to trimline and remove inner socket and plaster
6. Drill and tap inner socket for barb application
7. Screw in barb and seal with quick set two part epoxy and install tubing on barb
8. Drill hole in frame large enough to accommodate valve centered in the recess
9. Seal inner socket to frame by painting abraded inner socket with Orthocryl and inserting socket into frame. Finish sealing by drizzling Orthocryl into wrist opening
10. Feed tubing through valve hole and connect to valve
11. Seat valve into hole with quick-set two part epoxy
12. Seal in wrist unit of choice with Orthocryl

Fig. 1
Inner socket and outer frame prior to sealing

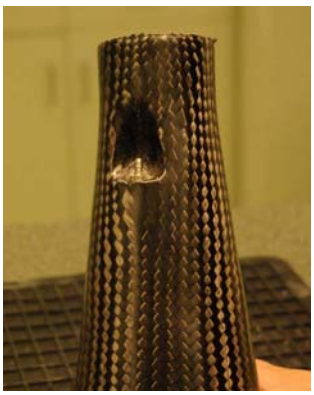

Fig. 2
Valve installed into recess built into frame

Fig. 3
Two prostheses prior to wrists unit installation

Application

The same liner used during casting is donned over the user's residuum. Again care must be taken to ensure no air gaps or voids are present between the liner and residuum. An air wick is applied over liner if one is not present in the liner. The socket is donned over residuum. The sealing sleeve previously applied to the prosthesis frame is rolled up to its full extent. The user

then applies distally directed force into the prosthesis. This will seat the residuum into the socket prior to applying vacuum. Vacuum is now created in the prosthesis using the hand pump.

Check-Out

Once the prosthesis has been donned and vacuum has been established the fit of the finished prosthesis is examined. A typical critique of the prosthesis fit and function should be made but the prosthetist should take special note of three areas.

Range of Motion: There will be some restriction to flexion due to the material from the liner and sealing sleeve in the anticubital fossa. Flexion should be no more than 10 degrees less than active flexion without prosthesis on. This would be an indication of an anterior trimline that is too high.

Comfort in distal distraction: When the prosthesis is pulled distally there should be no discomfort at the user's distal residuum. This will sometimes be described as "pulling" or "sucking". This is an indication of lack of contact between the residuum and socket prior to vacuum application, or a residuum that deforms significantly during flexion.

Reliability of vacuum over time: The vacuum should be tested at the time of initial application with the pressure gauge attached to the hand pump. The vacuum level should also be confirmed after 10 min. There should be no perceptible loss of vacuum during this time. Loss of vacuum indicates an air leak within the system.

Trouble Shooting Loss of Suction

The most common cause of air infiltration is the sealing sleeve. This is the least durable aspect of the system. Any small hole in the sleeve can allow air in and a reduction in vacuum over time. An incorrectly or incompletely sealed inner socket and frame can also allow air to leak between the layers of laminate. These leaks most often happen slowly and the suspension is still adequate enough to hold the prosthesis on but the many advantages of the vacuum are lost. In rare cases the valve itself can be faulty but this will result in the inability to achieve vacuum.

RESULTS

All eight patients were successfully fit with an elevated vacuum task specific prosthesis. All prostheses fit were examined and consistent vacuum levels were confirmed. None of the patients experienced any discomfort or skin trauma secondary to prosthesis wear. Exceptional suspension was noted in all cases. One patient discontinued use of prosthesis due to poor biceps musculature and a resulting range of motion limitation greater than 10 degrees. One patient required assistance in donning as the strength and dexterity of his contra-lateral hand was not sufficient for the donning process and manipulation of the hand pump used to initiate vacuum.

Each prosthesis was designed for a specific activity or activities which determined the patient's use of the prosthesis. Figure 4 shows a patient with a prosthesis created for sports such as baseball and golf. The patient was able to comfortably perform these activities with less restriction to motion. He noted improved control and performance. Figure 5 shows a patient with a prosthesis created for rock-climbing. The patient was able to suspend his entire body weight from the prosthesis without discomfort. Weightlifting, biking, and swimming specific devices were also fabricated with similar improvements in suspension and performance.

Fig. 4
Patient using elevated vacuum prosthesis for baseball

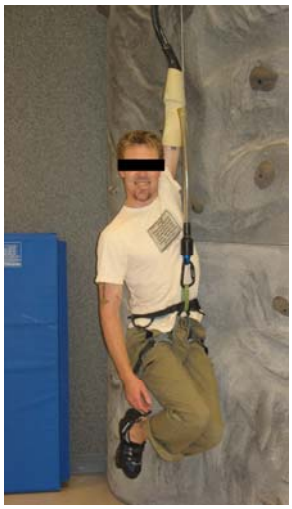

Fig. 5
Patient using elevated vacuum prosthesis for rock climbing. Patient's entire body weight is suspended from prosthesis

Note: harness worn by patient is for safety only and is not bearing any of patient's weight

FUTURE DEVELOPMENTAL FOCUS AND APPLICATIONS

The technique described in this paper explores the use of elevated vacuum for task specific prostheses only. The application of this technique to myoelectrically controlled prostheses is being examined. The hurdle of passing myoelectric signals through the silicone interface without sacrificing the elevated vacuum must be solved. As new materials and techniques are developed, many of which are currently being researched, it will be possible to apply elevated vacuum to a much larger range of prosthetic devices.

CONCLUSION

The eight patients fit with the elevated vacuum system have demonstrated a proof-of concept for the application of elevated vacuum in transradial activity specific devices. This procedure can be used as a springboard for future applications which will expand its clinical relevance.

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The Custom Silicone Interface: Clinical Applications and Observations

Robert Dodson, C.P.O., Bridget Jowid, O.T.R.

Abstract

Silicone interfaces have been utilized in the field of prosthetics in a multitude of applications over the last three decades. From suspension techniques to the enhancement of comfort and protection of atypical or diseased residual limbs, silicone has become an integral part of the prosthetic system. Beyond the field of prosthetics, the use of silicone has become a mainstay in wound management, scar maturation, and in the overall promotion of healing in the world of occupational and physical therapy. Extreme cases such as severe burns and or extensive skin grafts have forced many practitioners to consider utilizing silicone interfaces to protect fragile skin.

Clinically, as we incorporate custom silicone into the mainstream design of the "standard" upper extremity prosthesis, we are beginning to see unexpected benefits with the usage of this material on otherwise healthy skin. All of the therapeutic benefits of silicone are now being combined with normal prosthetic usage including limb healing and scar tissue maturation and management. This presentation will provide a historical look at the applications of silicone in the fields of prosthetics and occupational / physical therapy as well as consider unique applications in upper extremity prosthetics.

Introduction and History

Silicone, a synthetic polymer made of repeating silicon to oxygen bonds to form polymeric chains, has become a mainstay in medicine and is commonly utilized in a variety of medical applications. In 1824, Jons Jacob Berzelius discovered silicon from the reduction of silicon tetrafluoride with potassium. (Colas, Curtis, 2004). He unknowingly purified a substance that would change the world and unleashed one of the most widely used elements on the periodic table. Silicon is the second most abundant element on earth, eclipsed only by oxygen, and makes up 25.7% of the earth's crust although is not found free in nature and must be prepared for use in the industrial world (Lide, 2004). Frederick Kipping is credited as "the father of silicone chemistry" by developing a method for studying organo-silicon compounds. He published 54 papers on the subject between 1899 and 1937, but failed to see the potential commercial value of these new silicone compounds. In 1943, a new company by the name of Dow-Corning Corporation found an industrial use for silicone and began the manufacture of silicone polymers (Plastics Historical Society, 1986).

Silicone polymers have been used in many industries because of their exceptional physical properties ranging from liquids to hard solids (Colas, Curtis, 2004). Because of its unique biocompatibility and biodurability, silicone has found a permanent place in medicine through a wide variety of applications. Hydrophobicity, low surface tension, along with chemical and thermal stability are all properties that prompted silicone's initial introduction into the medical field. By the end of the 1960's, silicone materials were being employed in orthopedic applications, catheters, drains / shunts, kidney dialysis, heart bypass machines, and aesthetic implants (Colas, Curtis, 2004). Further development of silicone technology over the last four decades have branched into the rehabilitation arena in the form of scar management, wound care, burn care, and prosthetic interface materials.

Therapeutic Uses of Silicone

The use of silicone products in the therapeutic realm are largely related to the treatment and prevention of hypertrophic and keloid scar formation. These abnormal types of scarring can cause significant issues with range of motion, can be painful and may have psychological implications as a constant reminder of the trauma endured by the patient (Berman, 2007). In the late 1980's, various types of silicone elastomer sheeting began to be used on patients with these types of scar formations. Although the precise mechanism of action of silicone elastomer sheeting has not be clearly defined, there is evidence that consistent use of this product can help minimize scar formation and increase scar elasticity and has become widely accepted by occupational therapists in clinical practice (Berman, 2007).

Wound care and burn management have also seen the introduction of silicone materials into clinical practice. Burn management is mostly concerned with hypertrophic and keloid scar formation and the loss of range due to these changes in skin composition. There is some evidence that silicone gel sheets are more effective with the concomitant use of pressure dressings and that better results are found when worn under a pressure garment (deLinde,1992). This fact is supported by a study recently finished at the University of Taiwan in which the effects of pressure on the growth of human scar fibroblasts showed that with pressure application, fibroblast cell growth rate was slower than that of normal cells (Cheng, 2008). Whether or not this is a positive or a negative result needs further investigation.

Wound care is a therapeutic area that is beginning to see the introduction of silicone products in to the mainstream of clinical treatment. Products such as Biobrane dressing act as a semi-permeable "peseudoepithelim" which allows gas exchange at the wound surface. Evaporative water loss from the wound is decrease 90% and because of the adherence of the Biobrane to the wound surface, pain is reduced and bacteria proliferation is minimized (Smith, Price, 2002). The use of these materials requires that the wound itself be healthy. An infected or necrotic wound is not recommended for the application of silicone therapies (Hunter, 2002).

Prosthetic Uses of Silicone

The use of silicone in the discipline of prosthetics first began with the advent of the aesthetic hand prosthesis in the mid 1950s. The passive devices were used to restore near-normal appearance and improve a patient's overall function (Pillet, 2002). In 1986, Ossur Kristinsson introduced the first "silicone liner socket" in the form of the Icelandic Roll-On Silicone Socket. Over the past two decades, this concept and this technology has revolutionized the fitting of lower extremity prosthetic devices. Suspension and comfort were the main objectives of this type of silicone application and great success has been achieved by both clinicians and patients alike (Baars, Geertzen 2005). Through the use of silicone liner technology, new ideas are emerging with the application of vacuum assistance which eliminates much of the pistoning of the socket during swing phase by creating a negative pressure environment against the residual limb. Sealing type socket designs, first described by Haberman in 1995, utilize the flexible properties of silicone with the addition of a one-way valve to provide another option in the fitting of prosthetic devices (Haberman, 1995).

The recent emergence of the upper extremity specialist in prosthetics has led to a new focus on clinical applications of silicone materials and their benefit to the design of upper extremity prostheses. Dillingham showed that the average upper extremity amputation occurs in a traumatic scenario thus increasing the chance that these patients will have severe scarring on various areas of their residual limb (Dillingham, 2002). Working in conjunction with occupational therapists uniquely interested in upper extremity prosthetics, the advantages of applying silicone in scar, burn and wound management are now being correlated with benefits seen when silicone is introduced into prosthetic design. Uellendahl, etal. described the use of custom silicone sockets for myoelectric prostheses in 2006. This research concluded that custom silicone socket design enhanced three of the primary goals in fitting upper limb prostheses including comfort, function, and appearance (Uellendahl, 2006). Along with these enhancements, custom silicone sockets are now being seen as a way to promote skin health, provide a proper wound healing environment, and possibly reduce the negative effects of hypertrophic and keloid scar formation seen in patients with skin grafts and burns. Clinical observations are the first step in the research process and can begin the quest to answer basic questions of efficacy and necessity. Therapeutic benefits of silicone in the form of wound care and scar management can be seen in a recent case seen in our clinic.

Silicone Case Study

Mr. W. is a 43 year old white male that suffered third degree electrical burns to over 85% of his TBSA. This traumatic, on the job, injury in 1999 resulted in bilateral transradial amputations. He has severe scarring to his face, neck, chest, back, arms and legs. He fights MRSA infections on a regular basis which are primarily restricted to his facial region. He takes vitamins regularly and he eats two home cooked meals daily. “I am a healthy eater.” Over the past nine years, Mr. W. has worked with three different prosthetic companies and multiple occupational therapists. His initial prostheses were fabricated using inner flexible liners made of Proflex™ with silicone. This first set of bilateral prostheses could only be worn two to three hours per day and only when absolutely necessary. This was due in large part to the sensitivity and integrity of his skin when wearing the prostheses. A chronic wound developed on his left elbow and over the course of the last four years, many medical interventions including wound care and surgical debridement have been used in an attempt to heal this wound. Even with multiple courses of exhaustive methods using traditional wound healing therapies, the wound never fully healed. Mr. W. reports, “Whenever my left side would get pressure and moisture, a sore would develop.” “My wounds would bleed daily. I had to change my sheets on a daily basis.”

In June of 2007, Mr. W’s current prosthetist re-fabricated the inner flexible liner using a 10 shore silicone material. Within a month of using this new liner on a daily basis, Mr. W’s wound began to heal. He reports that the prosthesis was comfortable and that he was able to tolerate his prosthesis for 14-16 hours per day. Mr. W’s wound healing progress was followed over a four month period. By the end of the four month period his red weeping wound was healed with dry epithelial tissue. To this day, no further wounds have developed and Mr. W is wearing his prostheses daily for 14-16 hours per day.

This case study demonstrates the need for further case studies and research in on the implication and potential benefit of the use of silicone with persons with amputations, burns, and specifically for those persons with open wounds and/or scarring. These clinical observations raise the question how and why did a chronic wound heal inside a prosthesis, especially after traditional methods of wound healing were employed and failed. The opportunity for further research on this subject is great and should be considered.

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Upper Limb Prosthetic Outcome Measures (ULPOM) Group

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Background

In recent years there has been a sea change in the field of hand prosthetics, an increasing number of clinicians and researchers have a desire to be able to objectively measure the functional effectiveness of a prosthesis, or the ability of a user with their device. The problem has been that there are many tools to measure the function of hands and arms, but few seem appropriate to prosthetics. Also the tools that do exist seem to have conflicting aims and methods, so it is hard to choose the appropriate test. If practitioners have no meaningful way to test if a device is better for one user (compared with another device), they have no easy way to demonstrate to funders or providers that one solution is more effective than any other. Similarly, they lack a common language to simply pass on their professional judgement to their colleagues.

What does exist is an array of different tools for measuring different aspects of prosthetic design, function and use. There is little standardisation between centres in the same country, let alone across borders and seas. Worse, there is evidence that existing techniques are being invalidated (conceivably through ignorance, and definitely due to pressures of time to conduct a truly systematic study). To save time, or effort, fully validated tests are being shortened, or favoured sub tests are being selected from the greater whole, so that the results obtained are incomplete, invalid, or simply wrong.

Evolution of an approach

From the above, it became apparent to many in the field that there was a need to attempt to achieve greater knowledge and understanding by practitioners and engineers who are connected to this field.

The first stage in meeting this need was a workshop hosted by the Institute of Biomedical Engineering and chaired by Wendy Hill, (OT at the IBME limb centre), as part of the MEC Symposia in 2005. It aimed to advance the knowledge of outcome measures and allow sharing of some of the latest ideas and work in the field. This workshop was summarised by Virginia Wright in the Journal of Prosthetics and Orthotics [1]. The information contained in this paper fed directly, and indirectly, into subsequent moves to create a standard approach.

Following MEC, a small working group was set up to progress the ideas further. A workshop was hosted by Øyvind Stavdahl and the Norwegian University of Science and Technology (NTNU) with Wendy Hill and Peter Kyberd from the IBME as the organising committee.

It brought together an international group of researchers, therapists and users in the field of upper limb prosthetic research, to illuminate the technical outcomes measures from many perspectives. They attempted to identify key issues in the deployment of assessment techniques and tried to identify a more consistent means of making assessment of outcomes from prosthetics research and fitting.

It was stated that an ambitious long term goal sought by this work would be to:

Establish a terminology and methodology for the pre-clinical assessment and comparison of alternative prosthesis designs that allow for clinically relevant results to be reported and compared across different studies and countries.

Following the two-day workshop the results were brought back to a wider audience at the World Congress of the International Society of Prosthetics and Orthotics in Vancouver in August 2007. From this a broader working group, known as the Upper Limb Prosthetic Outcome Measures (ULPOM) Group was formed, organised by Shawn Swanson. The group's aim was to produce recommendations to help standardise the practice within the profession.

One of the implicit aims of the group was not to invent yet another test to add to the tests that already existed, so the process they have adopted has been to proceed through all the existing tests and identify those tests which are most appropriate to prosthetic use, to judge which of these are already validated for use with prostheses, or to identify what would need to be undertaken to validate them, and, in the long term, assist in the process of validating them. From this it is hoped that a toolbox of techniques and measures with a standardised approach to assessment will be evolved. This paper describes the methods by which this approach is being developed.

Assessment

The lifecycle of any healthcare product or device proceeds from the Research arena to Development and on through the Clinic to the Home. At each stage it is important to be able to measure the design against preceding designs, in order to gauge whether the changes are worthwhile and are what the designer intended. Each player in each area has a different perspective on what information they need or can use and a different way of seeing the problem. Often they have been unable to communicate unambiguously. This had lead to confusion about what a particular test has to offer and how it might perform the assessment. In 2002 the World Health Organization produced an International Classification on Functioning, Disability and Health (WHO-ICF) [2]. One consequence of this process was to express clearly the different information domains that exist in the development of a healthcare product or service.

The three domains are:

1/ Body structures and functions

2/ Activities
3/ Participation

They reflect the aspects of the provision and the mode of treatment/care required. Once expressed this way it becomes easier to see that each has its own set of particular requirements and so the process of assessing progress or function within each domain *has* to be different. It also becomes more apparent that previous attempts to measure function *without* such insight were less likely to prove effective, as the measurement might not be focused on the right domain, or using the right means to measure it.

The three domains map easily on to the three broad areas of assessment:

Functional Device performance - measures are related to the way the device works, such as speed of grip, strength, range of motion.

Activity Assessment of function within a clinical setting - Integrating grasp and hold, but using the device in abstraction.

Participation The user's experience with the prosthesis in everyday life; how the prosthesis is *actually* used.

Each area can then be measured using a different technique, shown table 1:

Domain	***Assessment technique***	**Methods**
Function	*Technical tasks*	Picking up objects, measuring grip strength, grasp types
Activity	*Clinical assessment*	Observations within the clinic or centre
Participation	*Self rating*	User's own opinion, questionnaires

Table 1 Assessment techniques and their relation to WHO-ICF domains

Naturally, each of these methods has their own biases and shortcomings. However, the most important conclusion to draw from this is that for a complete analysis, knowledge of each domain is necessary. One method can only serve some of the range. What is needed to unify them is not one test to cover all domains, but a single *approach*, with different tests to elucidate the information, hence the aim of the ULPOM.

A second insight is that through the lifecycle, different stakeholders have different levels of importance to the process as the different techniques dominate/operate. One way to express the relative importance is in figure 1.

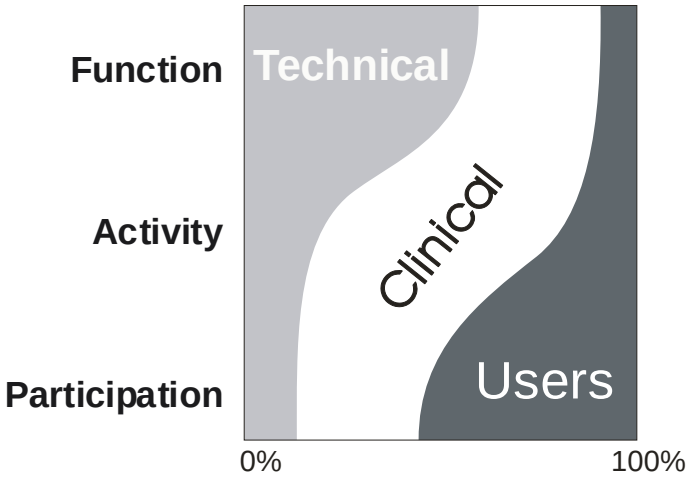

Figure 1 The relationship between the level of importance each stakeholder group has to the process of development of a prosthesis and the sorts of techniques needed to assess these domains

Thus, following the development of a prosthesis, initially: Initially the engineer wishes to know a prosthetic hand design can open wide enough to admit objects or close fast enough to be practical. S/he would then use basic *Functional/technical* tests, such as simple measurements based on motion tracking. Once the device is more complete, the assessment moves on to *Activity* based measures which are generally conducted in the clinic or lab. Can the device pick up household objects? Can it hold on to them and move them about? This information is important to the engineer, but now the input from the clinical team becomes important as their insight into its long- term use becomes relevant. Early fittings in the clinic also need activity- based measures as the clinicians need to compare the device to others, and to monitor progress in using a specific device.

Finally, the device moves out into the home and the activities may revolved around tasks specific to the needs of that user. This may tell the clinician about the functional capabilities of the device or of the person and something of their motivations, but when the therapist wants to know more about how the user feels about their device and how it integrates into their daily lives, then the information will more likely to be obtained through a questionnaire. Hence it can be seen that at each stage a different tool is used to obtain the information. Each is important and provides a different insight. Some techniques overlap into different domains, but only with multiple assessments will the full picture be clear.

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important and provides a different insight. Some techniques overlap into different domains, but only with multiple assessments will the full picture be clear.

From this, it can be seen that different existing tools cover different areas of the continuum, for example Figure 2 shows three tests straddling the four domains

Figure 2 Possible range of applicability of three different tests. SHAP is a technical test. ACMC is Activity based and COPMPUFI is a questionnaire.

An additional factor is the design of such tools. It tends to control how well it is received by practitioners, and if it is to be used generally. It must therefore be user friendly, easy to administer, inexpensive and the results must be easily understandable and interpretable. It must also have validity and reliability.. This ensures that the measures are repeatable and do not depend on who conducts the test or when or where.

Future directions

The ULPOM group is making progress towards the identification of the tests and tools that are appropriate and usable in the prosthetics arena. The process of dissemination of information has been conducted at MEC. Potential roadblocks that may hinder the progress towards effective outcome assessment have been identified and the group is addressing them over the next few years.

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ASSESSMENT OF CAPACITY FOR MYOELECTRIC CONTROL: CONSTRUCT VALIDITY AND RATING SCALE STRUCTURE

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INTRODUCTION

The *Assessment of Capacity for Myoelectric Control* (ACMC) is a 30-item standardized clinical assessment designed for the upper limb prosthesis group [1, 2]. It measures the quality of prosthetic hand movement performed by the prosthesis user during a self-chosen two-handed functional task. ACMC is suitable for prosthesis users of all ages and of all prosthetic sides/levels [2, 3] . Previously, repeated ACMC assessments of upper limb prosthesis users were used to evaluate the validity of the construct [1]. Since the strengths and weaknesses among these users were likely to be repeated several times in the data obtained, the abilities of the prosthesis users in that sample might not give the best picture of the functioning of the items. It was hypothesised that a wider range of ability across the sample might provide a better picture of the functioning of items. Therefore, a further validation of ACMC based on single measures was considered.

The performance of the 30 ACMC items is rated on a 4-point scale, ranging from 0-*not capable* – to 3-*spontaneously capable*. One concern is if the four ACMC categories are sufficient to differentiate the prosthesis users on the basis of their abilities. Another concern is whether the raters have used the four categories in the expected manner.

The overall aim of this study was therefore (a) to evaluate the construct validity of ACMC and (b) to examine the 4-point rating scale structure and its use. With a larger sample of single measures, specific questions were asked: Does a larger number of subjects provide a wider range of prosthetic ability than was found in the first validity study? Does the hierarchical order of ACMC items match the clinical knowledge about the difficulty of the items? Do all the items work together to measure a single "prosthetic control" dimension? Do all the items function as expected? Is the 4-point rating scale appropriately constructed to differentiate between prosthesis users with different abilities? Have the four rating-scale categories been used in the expected manner?

METHODS

Subjects, instrumentation and Procedure

Ninety-six upper limb prosthesis users with different prosthetic levels/sides/experience were included (males 58, females 38, congenital deficiency 83, amputation 13, age range 02-57 years, mean age 11, median age 8, experience 0 month – 19 years). Their ACMC assessments were collected by six qualified raters. The quality of prosthetic hand movement was assessed based on the 30 functional items that are grouped into four hand use areas:

gripping, holding, releasing and co-ordinating. All items are rated on a 4-point rating scale ranging from *0 - not capable* to *3 - spontaneously capable*.

Data analysis

Rasch analysis was used to analyse the data since it allowed an analysis at the assessment item and rating scale category level [4]. Rasch analysis is a mathematical technique for estimating linear logit (log-odds units) measures of item difficulty and person ability from ordinal data. The range of ability existed in the sample was examined by Rasch generated measure 'person separation index'. The construct validity was examined by the difficulty of the 30 prosthetic control items and the ACMC's dimension (by principal components analysis -PCA). Rasch generated fit statistics for items were used to find out whether any item deviated statistically from the expectation of the Rasch model, i.e. functioned unexpectedly.

The rating scale structure was examined by different Rasch generated parameters. 'Person ability measure' was used to examine whether each category represents a higher level of performance than the previous category. 'Threshold Measure' between every two adjacent categories was used to examine the effective use of 4-rating scale categories. If the threshold distance between any two adjacent categories is too large (> 5 logits) or too small (< 1.4 logits), then this suggests a need for an extra category or a collapse of two existing categories. Thirdly, 'outfit mean square' (MnSq) for each category is used to examine the consistency of use of the category. A category with outfit MnSq > 2.0 indicates that highly unexpected ratings are recorded in this category.

RESULTS

Range of ability in the sample

A wider range of prosthetic ability existed in this sample (person separation index 5.21) when compared to the first ACMC validity study (person separation index 3.79).

Dimensionality of ACMC

PCA confirmed that the data collected by the ACMC items is strongly unidimensional. This implied that all the items work consistently to measure the control of a prosthetic hand.

Difficulty and functioning of ACMC items

All 30-item difficulty measures and all 96 person ability measures are displayed graphically in the person-item map in Figure 1. On this map, the measurement scale is drawn vertically. The persons are plotted on the left side and the items are plotted on the right. The difficulty of ACMC items targeted well at the subjects' abilities. The mean item difficulty is set at zero logit (by convention). The mean person ability was +0.48 logit with SD of 2.81 logits. As seen in Figure 1, the two means are close together, indicating that the difficulty of items targeted well at the subjects' abilities.

The hierarchy of items shows that the items cover a substantial distance on the construct of prosthetic ability. Items relating to prosthetic hand movements performed without visual feedback are the most difficult items. Items that need good timing in catching or receiving objects are also relatively difficult. Prosthetic movements that are performed with the arm/hand supported are the easiest items. This hierarchy of item difficulty matches the clinical knowledge about the difficulty of different prosthetic hand movements.

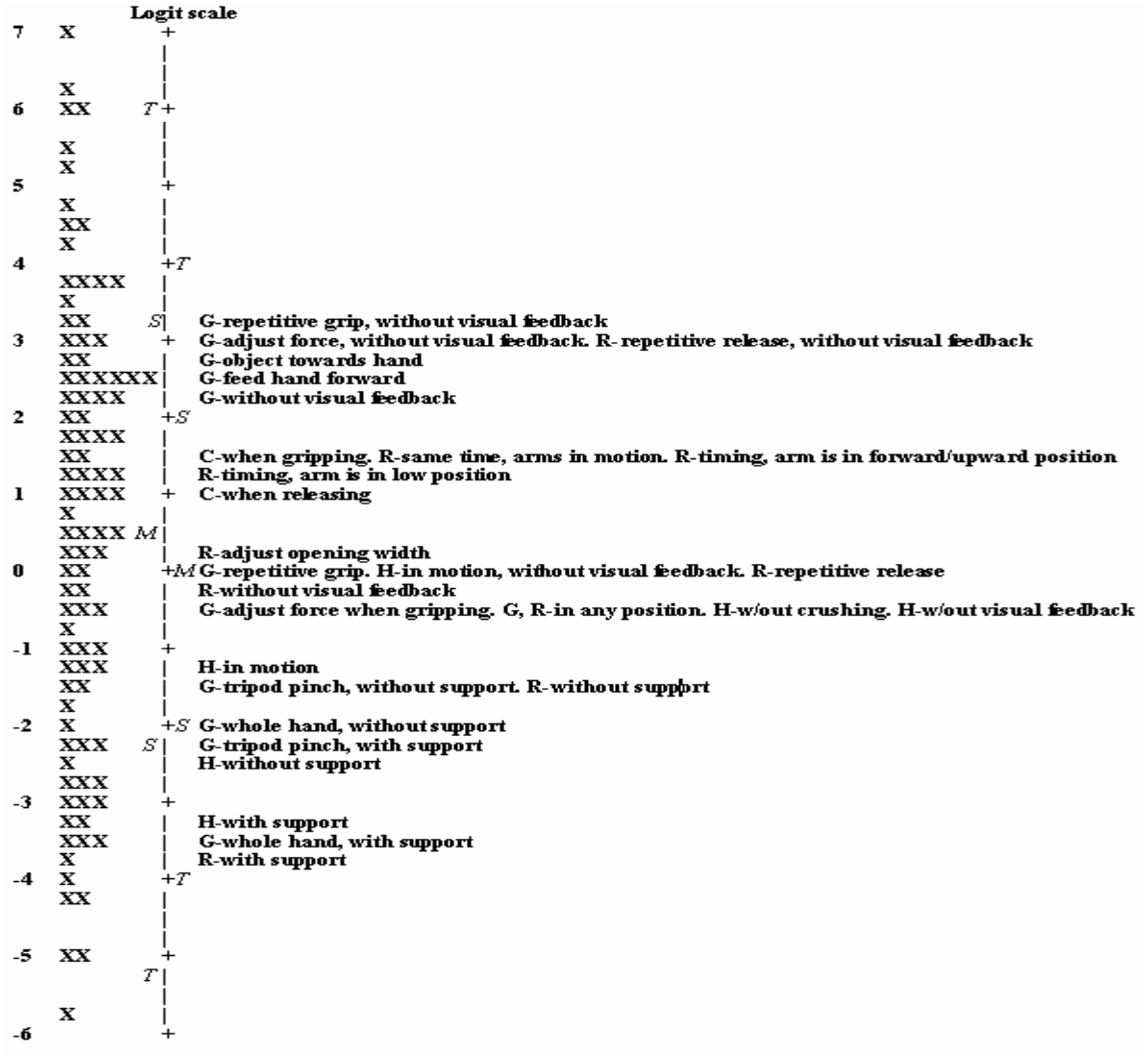

Fig. 1. Person-item map: person ability measures in relation to item difficulty measures
X= subjects, M= mean for person ability and item difficulty, S= 1 Standard deviation (SD) from the mean, T= 2 S.D. from mean, G= gripping, R= releasing, H= holding, C= co-ordinating, w/out= without

According to the fit statistics, two items functioned unexpectedly *(gripping-without visual feedback* and item *releasing –same time, arms in motion).* Three persons, who were rated higher than Rasch model's expectation, contributed to the 'poor functioning' of these two items. Both items functioned well after the removal these unexpected assessments. This suggested that the 'poor functioning' of items was idiosyncratic to this sample, not systematic to the items. It was therefore decided to retain the items in this analysis. The fit statistics of another item '*gripping - tripod pinch, with support'* showed that there was too little variation in the response pattern, perhaps suggesting that the item was redundant. Since a redundant item is no threat to the construct validity, this item will not be investigated further.

Rating scale structure and its use

Summary statistics for the four rating-scale categories are shown in Table I. The 'frequency of use' of categories 0, 1 and 2 were fairly even. Category 3 - *spontaneously capable* was used roughly three times more often than any of the other three categories (count = 949). This implied that many subjects were spontaneously capable in many items.

The distance between the 1st and the 3rd thresholds was 3.13 logits (from -1.72 to 1.41), indicating that the functional range of the rating for any particular item is about 4 logits. This was wider than the standard deviation of the person ability measures (2.81 logits), indicating that the 4-point rating scale usefully differentiates the persons on the basis of their abilities.

The person ability measure and threshold measure increases with the category value. The distance between two thresholds on each side of category 2 was 1.1 logits (1.41 to 0.31), slightly less than the 1.4 logits indicative of optimal rating-scale functioning, but large enough to be clinically unambiguous. The finding that the distance between thresholds on each side of category 1 was 2.03 logits (-1.72 to 0.31) suggests that an adjustment of the clinical criteria for category 2, "*capable on request*", to include a slightly lower level of performance, would improve the functioning of the rating scale.

Category	Frequency of Use	Threshold Measure	Observed Person Measure	Outfit MnSq
0 - not capable	388	NONE	-3.07	1.14
1 - sometimes capable	380	-1.72	-0.69	1.00
2 - capable on request	366	0.31	1.10	0.50
3 - spontaneously capable	949	1.41	3.90	1.09

Table I. Summary statistics for the four ACMC rating scale categories

- *Frequency of use: the number of persons rated in that category*
- *Threshold measure: The difficulty measure between every two adjacent categories*
- *Observed person measure: average person ability measure*
- *Outfit MnSq: This is used to examine the consistency of use of the category*

DISCUSSION

The consistency of item hierarchy with clinical expectation and the unidimensionality of ACMC confirmed the construct validity. The item misfit could be due to the difficulty of the tasks. The originators of another Rasch derived test found that persons' measures were dependent on the task performed during the assessments [5]. Thus, it is reasonable to assume that the control of prosthetic grip is easier in some tasks than in others. Therefore, further research is needed to test if the ACMC items are functioning in a similar way independent of the choice of tasks. Revision of category 2's definition – capable on request would improve the functioning of the rating scale.

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THE IMPLEMENTATION OF PROSTETIC UPPER EXTREMITY FUNCTIONAL INDEX (PUFI) IN FOLLOW-UP OF CHILDREN WITH UPPER LIMB REDUCTION DEFICIENCY (ULRD)

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Introduction

Over the past years, the importance of adequate assessment of arm and prosthetic functioning in children with ULRD has become clear. Research has shown that the use of standardized instruments adds relevant information on functioning of children with an ULRD. (1,2) The PUFI is one of the validated instruments that was developed to asses the functional status in children with prosthesis (3). Since 2004 we implemented the PUFI at the department of Rehabilitation medicine of the Erasmus University Medical Centre and more recently, at the University Medical Centre Groningen as a standard procedure to monitor functional outcome and prosthetic management of children with ULRD. This prospective study is an evaluation of the data collected so far.

Method

Patients

All children visiting the outpatient department of rehabilitation medicine of our centers fill out the PUFI-questionnaire when they have an appointment once a year. The PUFI results are printed and are discussed during the visit at the attending physician.

We assessed 40 children (23 girls, 17 boys), 27 children in Rotterdam, 13 in Groningen with the PUFI at the mean age of 9.8 years (SD 4,0 years, range 8.3 years), all between 4 and 18 years. Two children had an acquired deficiency; the others had a congenital deficiency of the upper limb. Almost all children had a deficiency distal from the elbow and proximal from the wrist, one had al deficiency above elbow, another one had a bilateral deficiency. Fifteen children had a deficiency on the right side, 24 deficiencies were on the left side.

Instrument

The PUFI (Prosthetic Upper extremity Functional Index) evaluates the extent to which a child actually uses the prosthetic limb for daily activities, the comparative ease of task performance with and without the prosthesis, and its perceived usefulness, which are respectively scored on a 6-point nominal scale, 5-point ordinal scale and 3-point ordinal scale. Higher scores represent more ease of performance and higher usefulness of the prostheses. Sum scores range from 0 to 100. In addition, we calculated an adapted score in which only those activities are scored, for which the prosthesis is actually used. In previous studies, the PUFI showed good validity and test-retest reliability. (1,2)

There are two versions: the young child version, (age 3-6 years), containing 26 items, and the older child (age $\geq$ 7 years) with 38 items, which have 14 activities in common. In our department, in children up to 12 years old, the parents and the child fill out the older child version. Fifteen parents completed the young child-version, 25 children were assessed with the older child version, and in 5 cases the test was also completed by one of the parents. Fourteen children had more than one assessment in time.

Statistical analysis

For cross-sectional analysis, the last completed assessment of a child was used. In older child assessments the child's evaluation was used, also when a parent's version was available.

We compared the performance of activities for children who wear a prosthesis (users) with those of children who do not wear a prosthesis (non-users) using the t-test for independent samples. A t-test for dependent samples was used to compare the performance of children who have a prosthesis, tested for activities with and without their device. P-values ≤ 0.05 were considered significant.

Results

Use of prosthesis

Seventeen of 40 children do not use a prosthesis (non-users), of the 23 children using a prosthesis (users). Twenty-five wore a myoelectric prosthesis, 5 a passive (cosmetic) prosthesis. Of the prosthetic users 5 wore their prosthesis 0-2 hours/day during weekdays.

Method of performance.

In users and non-users bimanual activities are performed independently, users/nonusers need help in 2/3% and are not able to do the activity in 4/1%. Users can use their prosthesis active or passive in 51% of all activities. There is a significant difference ($p<0.000$) between users and non-users in performing activities with the residual limb. One-handed performance of activities did not differ between users and non-users.

Table 1: method of use prosthesis or residual limb.

Method of performance % of activities	users n=23, mean (SD)	non-users n=17, mean (SD)
prosthesis actively	22 (21)	0 (0)
prosthesis passively	29 (22)	0 (0)
residual limb	34 (28)	88 (10)*
one-handed	10 (12)	8 (9)
some help	2 (3)	3 (3)
cannot do	4 (9)	1 (2)

* Significant difference between users and non-users

Type of activities

In the young child-group, the prosthesis was frequently used for almost all activities. Top 5 includes cycling, eat raisins, open juice pack, climb a slide. Older children reported cycling as an activity in which they used their prosthesis most often, whereas parents hardly reported this activity. Other favorites include: draw a line, open a pencil case, and open a bag of crisps.

Ease of performance

For prosthetic users overall scores on ease of performance were moderate, mean 71.3 (SD 12.6). For only those activities for which the prosthesis was actually used scores on ease of performance were significantly better 86.8 (SD 8.8) (p=0.000). A significant difference was found for users doing activities with prosthesis compared to performance without prosthesis (p=0.029). Comparing the performance of users with non-users, users seemed to have lower scores on overall performance (p= 0.002). However, specific performance of activities for which the prosthesis is used, is comparable between both groups (mean score 86.8 versus 89.9)

Table 3: Ease of performance of functional activities.

	Mean	Differences non-users/users without prosthesis	Differences non-users/users with prosthesis	Differences users with prosthesis/without prosthesis
Ease of performance				
Non-users	89.9	*p*=0.041*	*P*=0.002*	
Users - without prosthesis	82.8			*p*=0.029*
Users - with prosthesis	71.3			
Ease, specific activities				
Users - with prosthesis	86.8		*†p*=0.227	*†p*=0.195

** Significant differences between groups*
† Differences calculated for users with prosthesis for specific activities

Usefulness
Users perceived their prosthesis useful, mean score was 53.8 (SD 23.3); in specific activities in which they used their prosthesis active or passively usefulness is significantly higher, 78.2 (SD 16.1) ($p<0.000$).
Follow up
Eleven children had two follow-up measurements, 3 children had three, with a mean interval of 1,3 years between the assessments.
Overall, there was a tendency of improved performance over time in the children with prosthesis. The perceived usefulness of the prosthesis for specific activities improved 5 to 10 points in 3 children, 4 children improved 15 to 20 points, and 2 improved more than 30 points. Two children did not change; one child scored about 5 points, 2 about 10 points lower.

Discussion
In previous studies we demonstrated that users (with and without prosthesis) and non-users perform good in functional activities. The present study confirmed that children tend to use their prosthesis for specific activities. In our rehabilitation centers, children and parents get extensive information about the possible benefits of a prosthesis. This gives insight whether prosthesis could be useful, and if so, what type of prosthesis would best fit their individual need. This approach is based on the current practice in our departments, that that possibly not all children need to wear prostheses and that a prosthesis can be beneficial for specific activities, or for a broad range of activities. (4)

The PUFI can be used to evaluate performance and usefulness in users of a prosthesis both on an individual level, and in a larger cohort.
This project contributed to the development of a network with other rehabilitation centers in the Netherlands aiming to combine our data and hereby increase our knowledge on the functioning of children with ULRD. (5) Although the PUFI is developed for use in children with prosthesis, we find part of the PUFI also useful for evaluation of children without prosthesis, especially since no other functional performance measures are available. We look forward to current development of the UFI for the non-prosthetic group.

Conclusion

The study showed that assessing functional activities in children with ULRD using the PUFI provides useful information to monitor patients and to assist clinical judgement and adequate goal setting. Also, the results suggested that the PUFI is capable of measuring change over time in an individual child.

Acknowledgements

We would like to thank all children and parents who participated in this study, and Rehabilitation Center Maartenskliniek at Nijmegen for contributing.

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DO CHILDREN AND YOUNG ADULTS WITH UPPER LIMB REDUCTION DEFICIENCY PERFORM DIFFERENTLY?

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INTRODUCTION

Since 1999 the arm prosthetics team of the Erasmus University Medical Center has been involved in development, validation and implementation of measures of functional status in children and adults with an upper limb reduction deficiency (ULRD) (1,5,6) Internationally now validated instruments such as the Prosthetic Upper Extremity Functional Index (PUFI) and Unilateral Below Elbow Test (UBET) are available (2,3,4) The clinical use of these measurement instruments and interpretation of the results is still a matter of discussion (7,8) We do not yet understand why patients choose to wear or not to wear arm prostheses, nor can we predict the amount and pattern of use. From previous research we know that both wearers and non-wearers perform well in activities of daily life. Still 30-50% of the patients with ULRD do choose to wear an arm prosthesis. Arm prostheses seem to have functional merits for specific activities. In addition they are likely to have a considerable personal function, in which self-esteem and cosmetics play important roles as well. The importance and influence of these aspects have not yet been evaluated. No longitudinal cohort studies or intervention studies have been done until now, so we do not yet know how capacity of prosthetic use and performance in daily life change with age.
Assessing aspects of functioning on the different ICF levels and searching for relationships among them may lead to better understanding of the functionality of arm prostheses and the performance of patients with ULRD.
In this paper we will explore whether age might be a relevant factor for functioning with or without a prosthesis. The first purpose is to compare two age groups regarding capacity and performance of functional activities with or without prosthesis, measured by the PUFI and UBET (2,3,4) The second purpose is to get further insight in clinical usefulness of these instruments.

METHODS

Patients

In this study two age groups participated, 20 children and 20 young adults recruited from the Department of Rehabilitation Medicine of Erasmus University Medical Center Rotterdam (Table 1), all with a one-sided below elbow congenital reduction defect.

Table 1

	N Male	N Female	Mean age (SD)	Non Wearers	Wearers	Myo	Passive
Children	10	10	8.7 (2.0)	11	9	8	1
Young Adults	9	11	19.4 (4.7)	12	8	5	3

Procedures and measurement

Each participant had one visit to the Departments of Rehabilitation Medicine or was visited at home. Each participant performed the Unilateral Below Elbow Test (UBET). For those using a prosthesis, the UBET was first performed with the prosthesis followed by test performance

without using the prosthesis. In addition, each participant or his/her parent filled out the Prosthetic Upper extremity Functional Index (PUFI). Wearing time, method of use (capacity and performance), ease of performance and usefulness of the prosthesis were evaluated. In addition adapted scores for ease of performance and usefulness were computed in which only those activities are scored, for which the prosthesis is actually used.

Results

Both age groups seem to differ in wearing time, we saw an increases of wearing time with age.

Table 1 **Wearing time**

	0-2 hrs/day	3-5 hrs/day	6-10 hrs/day	11-15 hrs/day
Child	3	3	2	1
Young Adult		1	3	4

Users of a prosthesis can do more than they actually perform in daily life, so there is a difference between capacity and performance. This discrepancy seems to be larger in the young adult group. (Table 2 and 3, the sum of active and passive use in UBET versus PUFI for young adults 84% versus 34%, children 68% versus 30%) The prosthesis was used for different activities in both groups. The children used it more for playing activities, the young adults more for activities of daily life. Both groups used the prosthesis for riding bicycles. The method of use of the prosthesis varied between the groups (table 2,3) The young adults tend to use the prosthesis more in an active way and use their residual limb less than the children. More striking was the difference in use of the residual limb in the non-user group and users without prosthesis group (table 2 UBET) The young adults act more passive with their residual limb compared to the children who make more active use of the residual limb and use their elbow/trunk. The non-user children seem to act far less one-handed than the children without the prosthesis and the young adults on performance measures (Table 3 PUFI)

Table 2 **method of use UBET (capacity)**

Method of use prosthesis/ residual limb (% of activities)	Young adult Users with prosthesis (n = 8) Mean (SD)	Child Users with prosthesis (n = 9) Mean (SD)	Young adult Users without P (n = 8) Mean (SD)	Child Users without P (n = 9) Mean (SD)	Young adult Non-users (n = 12) Mean (SD)	Child Non-users (n = 11) Mean (SD)
Actively	35 (26)	30 (28)	19 (28)	44 (24)	18 (21)	55 (30)
Passively	49 (33)	38 (25)	53 (35)	19 (15)	56 (32)	17 (13)
Elbow/trunk	0 (0)	11 (14)	8 (14)	16 (17)	9 (16)	27 (27)
One-handed	14 (33)	21 (13)	18 (34)	21 (23)	17 (33)	1 (3)

Table 3 **Method of use of prosthesis or residual limb in PUFI (performance)**

Method of performance (% of activities)	Young adult Users with (n = 8) Mean (SD)	Children Users with (n = 9) Mean (SD)	Young adult Non-users (n = 12) Mean (SD)	Children Non-users (n = 11) Mean (SD)
Prosthesis actively	20 (29)	15 (18)	0 (0)	0 (0)
Prosthesis passively	14 (16)	15 (21)	0 (0)	0 (0)
Residual limb	33 (26)	41 (34)	77 (12)	85 (12)
One-handed	27 (16)	23 (18)	19 (11)	5 (6)
Some help	2 (2)	2 (4)	3 (5)	3 (3)
Cannot do	5 (5)	4 (4)	1 (2)	7 (9)

The young adults have higher scores for ease of performance compared to the children. Users with prosthesis of both groups have lower ease of performance scores than the non-users. However, the adapted ease of performance score over only those activities for which they really use their prosthesis is much higher and comparable to the scores of the non-user group. (Table 4)

Table 4 **Ease of performance of functional activities**

	Child Mean	Child Min	Child Max	Young adult mean	Young adult Min	Young adult Max
Non users	79.9	54.6	95	95.3	82.9	98.7
Users without P	75.8	56.8	88	87.3	78.5	94.1
Users with P	45.2	5.6	75	60.1	11.3	91.5
Users with P Specific activities	88.0	66.7	100	95.9	86.3	100

Usefulness scores seem higher in the young adult group. Again the adapted score for usefulness was higher in both groups.

Table 5 Usefulness

	PUFI Mean (SD) N = 8	activities		PUFI Median (range) N = 9	activities
Young adult	43.8 (9.7-81.8)	all	Child	37.5 (1.9-58.8)	all
Young adult Specific act	86.4 (71.4-100)	riding a bike, pushing lawnmower and twisting lid off a bottle.	Child Specific act	75 (66.7-100)	Riding bike, scissors

Discussion

In this study we found differences between age groups of patients with ULRD in wearing time, method of use and performance. Both age groups perform well with and without prosthesis. Both children and young adult users seem to have slightly more difficulty in performing activities without their prosthesis than the non-users, as seen in other studies (7) Non-user children act less one-handed than the other groups of children and young adults. An explanation might be that these children make more functional use of the residual limb.
The discrepancy between capacity and performance in daily life possibly grows with age. One would expect that prosthetic skills, wearing time and motivation increase with age and that ease of performance therefore is better in young adults. This is indeed shown in his study. The fact that the discrepancy between capacity and performance was larger in the young adult group compared to the children was an unexpected feature. Everyday use apparently does not develop while ease of performance and usefulness scores do. Personal and environmental influences, such as developing self-esteem and the importance of cosmetics may also influence the method and amount of use of the prosthesis in the group young adults. Additional measurement of skills of prosthetic use and personal and environmental influences may give a more clear view at the relationships of prosthetic use and performance during development.

Longitudinal standardized follow up of capacity and performance of patients with ULRD seems interesting and clinical useful according to our study. For a better comparison of the performance of the prosthetic users to the non-user group we advice using the PUFI adapted scores for ease of performance and usefulness assessment.

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A STUDY OF THE USE OF COMPENSATION MOTIONS WHEN USING PROSTHETIC WRISTS

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INTRODUCTION

It is well known that the functional capability of a prosthetic hand is less than that of the natural hand and thus it cannot perform the majority of the tasks a natural hand does, as well or as easily. Of the many consequences from this circumstance is that the prosthetic hand is generally used in a support role when there is a contralateral natural hand available. This is because the prosthesis is unable to grasp objects as flexibly as the natural hand. It is known, from splinting studies of the wrists of unimpaired volunteers, that if the person is not able to present the hand in the correct orientation, even the most flexible hand cannot perform prehensile tasks easily, [1,2]. In the prosthetic circumstance, without a wrist to orientate the hand relative to the object, conventional terminal devices do not grasp as effectively. The user must move their arm in a different way, that allows the prosthesis to be presented to the target in an orientation that will facilitate a secure grasp. A result from this is that there is a very real risk that these compensatory motions use greater ranges of motion, larger forces, or occur more often than necessary with a natural hand. Kidd et al [3], observed that these are three of the conditions likely to induce the changes in the musculoskeletal system that are referred to as injuries of overuse. There are few long term studies of the effect of overuse in prosthesis wearers. It is a well known observation amongst the clinicians that users who do not use their prostheses tend to suffer from the sort of degenerative changes associated with overuse. Less still is known about the effect of the compensatory actions of the contralateral limb.

Ross et al [4,5] have begun to address this absence in the paediatric population, by studying the motions of the users of two forms of wrist, using activities of daily living and a three dimensional motion tracking system (Vicon). The work described here aims to perform a similar study in the adult population. It is clear that if the tasks performed are to be meaningful they have to be selected to match the skills and knowledge of the potential subjects. Thus paediatric tasks must be at the appropriate cognitive level for the subjects, and for any group the tasks must be culturally appropriate. While Ross's tasks aim to engage children below the age of 12, such tasks cannot be expected to be suitable for an adult population. Hence the purpose of this study was to develop a series of tests that can be used by an adult population.

DESIGN

Within the World Health Organization's International Classification on Functioning, Disability and Health (WHO ICF), [6], this assessment is within the *Activity*

domain, straddling as it does the areas of *Research* and *Clinical Application*. The two areas have subtly different requirements. If the actions of the assessment are very closely constrained then any deviation from the general population will be easily seen and quantified, but the activity runs the risk of being seen as not being clinically relevant. A more relevant action may be too poorly specified to be reproducible and thus not easily compared test to test, or subject to subject. The method proposed here, aimed to be a tool that would be extended towards clinical assessment. The choice was made that the tasks should be selected to encourage natural bilateral motions. It was anticipated that the prosthesis would be used in a support role, but the tasks would make it necessary to use the prosthesis to complete the task.

In order to make the test manageable, six tasks were chosen from a list of sixteen already employed by Stavdahl in his design of a novel wrist orientation [7]. The selection was made to choose easily achieved tasks that did not create redundant data. This was based on experience of the designers and the use of the tasks in the earlier study.

The tasks were:

1. Hanging clothes on a clothesline with clothes pegs
2. Slicing bread
3. Eating with a knife and a fork
4. Sweeping the floor with a broom
5. Stirring in a pot
6. Cutting with scissors

The subjects were tracked using an eight camera Vicon system. Each participant was marked with 25 markers of different sizes on the head shoulders and arms. They were then recorded performing the tasks and the data processed to generate information concerning the ranges of motion and other kinematic information.

Subjects

Twenty unimpaired subjects were recruited to create a baseline understanding of the tasks. They performed the tasks both normally and impaired, using splints to simulate the use of a prosthetic hand and olecranon suspension socket. Results of the splint study were presented in [8]. Additionally, four prosthesis users were recruited and they performed the tasks using their prostheses. Each subject was instructed to perform the tasks, but no guidance was given as to how they should perform the task so that the resulting actions were spontaneous.

RESULTS

As anticipated, the different tasks resulted in different uses of the prostheses during each activity, allowing the use of the devices to be studied. The tasks fell into three groups: Those where the hands were used independent of each other so that their motions were only constrained by the task (eating and slicing bread); those where the hands needed to co-ordinate to manipulate the tools used (Hanging, Cutting,

Stirring); and the last where both hands were on the broom and so they moved in complete synchrony with each other.

The kinematic results showed that there were differences in the motions and angles used by the subjects. One of the most striking actions was performed by a prosthesis user. It was not uncommon to observe that when required to cut a circle in a piece of paper, many subjects used the scissors to hold the paper, when readjusting the position of the hand to grasp and reorient the paper. However one subject then changed direction of cutting, and instead of continuing in the same anticlockwise direction from the top of the circle, she started cutting in the *opposite* direction from the *bottom* of the circle, reducing the need to change the orientation of the paper and so the number of adjustments required.

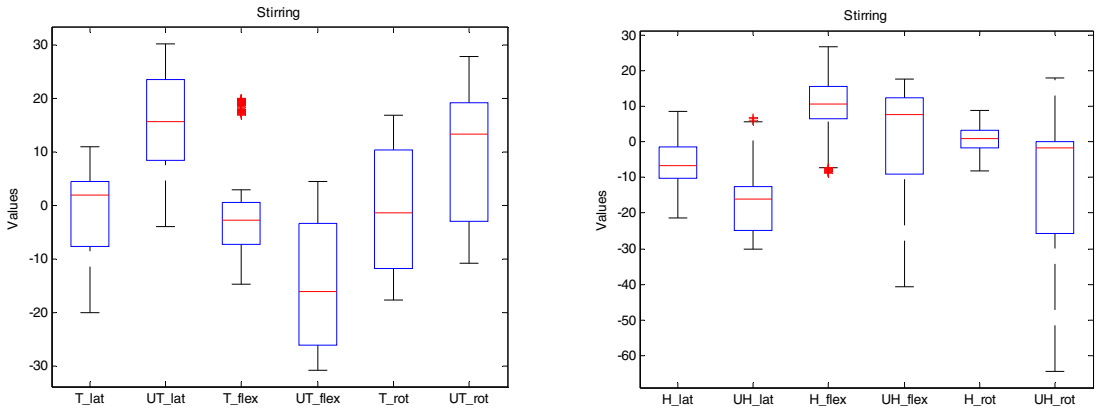

Figure 1 Ranges of motion recorded for both groups
T = Trunk to the left, H = Humerus to the right, , U = Users to the left

Each task has a different emphasis in the use of the prosthesis, the stirring task is chosen here as being a task that requires the two limbs to be loosely connected through the bowl, but it also allows some variation in grasping and use of the tools and prosthesis. Figure 1 shows the recorded ranges of motion for the trunk, and the humerus for the unimpaired side. It can be observed that the compensations are throughout the upper body. The trunk is laterally tilted away from the task in order to allow the restricted movement of the prosthetic side to be accommodated. Similarly the user has to lean forward. The position of the humerus is closer to the unimpaired subjects, but there is a far wider range of motion to compensate for the lack of any stirring performed by the wrist.

DISCUSSION

The level of constraint on the motion of the prosthesis needed to perform a task is an important one when considering the tasks. The constraints mean that greater demands are placed on the arms to position the device in the correct position.

The flaw in not constraining the actions more is that the ranges of motions can thus be as wide as possible. Though the subject who cut both directions round the circle appeared to have found a novel solution to save herself time and effort, a review

of the other subjects showed that one of the unimpaired subjects *also* cut this way. Thus it cannot be seen as a diagnostic indication of the difficulty of prosthesis use.

The need for increased range of motion for all the remaining joints of the arms and torso of the users was easily observed. This shows that it is not merely the arm that has the impairment, but the rest of the body that has to compensate for the loss of functional range. Hence when looking for the effect of limb loss or prosthetic design it is important to observe the entire body in detail.

The future direction of this research is to refine the tasks in order to select tasks that naturally constrain the motion and do not allow for too much individual variation, without needing to dictate to the subjects.

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Translation and linguistic validation of the Swedish version of "Orthotics and Prosthetics Users' Survey" (OPUS)

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Abstract:
There is a lack of valid Swedish instruments to assess the overall outcome of upper limb prosthetic treatment. The Orthotic and Prosthetic Users' Survey (OPUS) consists of five questionnaires assessing common goals in prosthetic and orthotic practice. The OPUS measures health related quality of life, satisfaction with device and services, respectively, and functional status of upper and lower extremities, respectively. Hence, this instrument could be a useful tool for outcome assessment of Swedish practice. Following the guidelines by the World Health Organization, the questionnaires were translated to Swedish and validated linguistically. Thirty-nine persons (12 men, 27 women) representing the target groups for OPUS participated in the study. During a regular visit at the prosthetic-orthotic out-patient clinic, the participants answered the relevant questionnaires and were systematically debriefed immediately afterwards. In most cases the items were understood as intended. In a few cases words and expressions had to be changed to avoid misunderstandings or diverse interpretations of the items. The resulting Swedish version of OPUS showed acceptable linguistic validity. A study on construct validity and reliability of the Swedish OPUS is in process and preliminary results will be presented.

LEARNING TO USE AN UPPER ARM PROSTHESIS: DOES ORDER OF PRACTICE MATTER?

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INTRODUCTION

Although most people are fitted with a prosthesis after upper extremity amputation, the prosthesis is often not used in daily life. In fact, 20 to 40% of the upper extremity amputees do not use their prosthesis at all, mainly due to a low degree of functional use [1,2,3]. This functional use can be enhanced by training [4,5]. For example, Carter et al. [5] showed that of the amputees who received training, 90% used their prosthesis in daily life, compared to only 50% of the amputees who did not receive a training after fitted with a prosthesis. Thus, training helps, but how should a training look like?

The current training methods used in the rehabilitation of upper limb amputees seem effective considering the fact that amputees learn to handle their prosthesis. However, there seems to be room for improvement because the prosthesis is not used in tasks that have been trained [6].

The current study examined the role of the structure of the training in the effect of prosthetic training to determine which order of practice tasks (i.e., random or blocked) had the highest effect on performance with an upper limb prosthesis. We examined training of able-bodied participants that used prosthetic simulators because there are only few upper limb amputees. We used two types of simulators, a myo-electric prosthesis and a body powered prosthesis with a voluntary opening hand, because these are the types of prostheses most widely used.

METHODS

72 able-bodied participants (36 men and 36 women (sd) age 21.07 (2.32) years) used an upper arm prosthetic simulator that strongly resembled the functioning and control of a real prosthesis (Figure 1); 36 participants used a myo-electric simulator and 36 participants used a body-powered simulator. For each simulator there were four groups of participants: group one practiced random and was tested random (RR), group two practiced random and was tested blocked (RB), group three practiced blocked and was tested blocked (BB), and group four practiced blocked and was tested random (BR).

On the first day, in the acquisition phase, participants received a training in which three tasks had to be executed, each consisting of 20 trials. To determine the effect of learning from the first day, a retention test—with execution of 5 trials of each training task—and a transfer test –with 5 trials of three new, more functional based tasks—were conducted on the second day. The three tasks to be executed in each phase were based on direct grasping, indirect grasping and fixating [7].

In the acquisition and the retention test a wooden cylinder had to be grasped with the prosthetic hand (direct grasping), a cylinder had to be handed over from the sound hand to the prosthetic hand (indirect grasping) and a ruler had to be fixated to draw a straight line (fixating). In the transfer test a mug had to be placed on a shelf by the prosthesis, a jar had to be handed over to the prosthetic hand after which the lit had to be turned off, and for the fixating task a pencil was sharpened with a sharpener. Initiation time—time from the start signal to initiation of the movement—and movement time—time between start of the movement and end of the movement—were used as outcome measures, recorded in milliseconds.

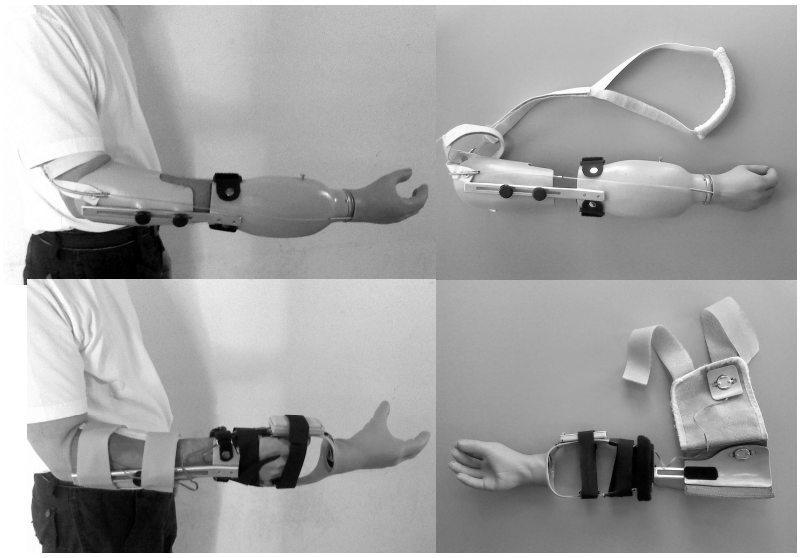

Figure 1. The body-powered simulator (upper row) and the myo-electric simulator (lower row)

RESULTS

We found no effects of initiation time. Although movement times of both the random and the blocked group got faster during the acquisition (p = .000), the blocked group had faster movement times in the acquisition phase than the random group (p = .009) and learning in this group extended over the complete acquisition phase (p = .000). However, this advantage disappeared in the retention and transfer tests, and no differences were found between the groups in the tests. This can be seen in Figure 2. Another interesting result was that the movement times were faster with the body-powered simulator in acquisition (p = .004) and in the transfer test (p = .034) compared to the myo-electric simulator.

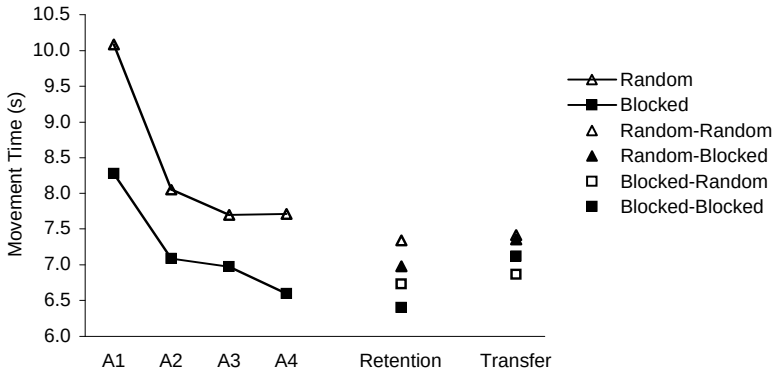

Figure 2. Movement time in seconds for each of the two groups (random (R), blocked (B)) in the acquisition (divided into 4 blocks) and for the four groups (RR, RB, BR and BB) in the retention test and transfer test.

CONCLUSION

The fact that no differences were found between the groups in the retention and transfer tests, implies that the order of practice does not influence performance after training or the performance of tasks other than trained. The clinical implication of this finding is that neither of the two tested orders of practice tasks (random or blocked) is preferred over the other. However, the results did show that training in a blocked order leads to faster performance.

Therefore we suggest practicing at least a part of the training in a blocked fashion. The advantage of a blocked order is that people learn more quickly how to handle the prosthesis, which saves time prosthetic learners have to spend in rehabilitation. This steady improvement can motivate the trainees to pursue the training and use their prosthesis more often at home.

NOTE

A full version of this paper is accepted for publication in the 'Archives of Physical Medicine and Rehabilitation'.

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Compliant Hands: The Next Evolution of the Prosthetic Hand

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While the last 10 years have brought significant advancements to the upper extremity prosthetic patient population, the function of prosthetic terminal devices has been limited by their inability to mimic the myriad grasping patterns of the human hand. Recently, several groups have made considerable progress in the development of terminal devices that offer more anatomical articulation while retaining the aesthetics of a human hand. The earliest generation of compliant hands expands function beyond the basic 3-point grip using the 1st, 2nd, and 3rd digits to a grasping capability that utilizes all digits in multiple, user-selected configurations. This technology provides a new level of stability and dexterity when handling asymmetrical objects of various sizes. Later generations of complaint hands envisage independent finger motion with sensory input capacity. The most challenging aspect in the development of an advanced compliant hand is in accessing sufficient data from the residual limb to control the increased degrees of freedom. This presentation will examine the current state of compliant hands and the direction of technological research and development.

DEVELOPMENT OF A CLINICALLY VIABLE MULTIFUNCTIONAL HAND PROSTHESIS

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1.0 INTRODUCTION

We have developed a new multifunctional hand mechanism in the hopes of providing a new mechanism that will have superior function over today's single degree-of-freedom (DOF) mechanisms and yet be robust enough to be clinically viable. There have been a multitude of multifunctional hands built, all of which have failed to find clinical application as an artificial hand replacement. Bolstered by advancements in motor and robotic technology, the past two decades has seen significant effort, and money invested in the development of externally-powered multi-functional hand systems [4, 7,8,9,10,11]. However while many of today's commercially available externally-powered systems owe their origins to modern upper-limb research, save for the Touch Bionics hand [14], no multifunctional hand mechanisms have made the transition from the Laboratory into clinical practice.

In the end, most multifunctional hand designs are doomed by practicality, even before the control interface has become an issue. Prosthesis users are not gentle with their devices; they expect them to work in all sorts of situations never dreamed of by their designers. Most mechanisms fail because of poor durability, lack of performance and complicated control. A multifunctional design is by its nature more complex than a single DOF counterpart. Articulated joints on fingers are more likely to fail than monocoque, or solid finger designs. To be clinically viable, the fingers and thumb of most prosthetic hands are non-articulated and have a single axis of rotation. This minimizes the number of moving parts, reduces complexity and increases robustness. Palmar prehension is achieved by single joint fingers that are fixed in slight flexion at a position approximating the interphalangeal joint. The resulting finger shape also creates a concave inner prehension surface that can be used to provide cylindrical prehension. Another practical consideration is performance. The hand must be able to generate enough torque and speed, and have a sufficient width-of-opening to be useful to the user [6]. The pinch force of a multifunctional hand does not have to be as high as that of current commercially available single DOF hands because of the adaptive nature of their grip. But they should still be capable of high speeds-of-opening and have a pinch force of at least 68 N (15 lbs_f) in accordance with Peizer et al [13].

A possible compromise to the dilemma of robustness versus increased function is to limit the device to those degrees-of-freedom necessary to replicate Keller et al.'s [1] grasp patterns. This idea of providing sufficient DOFs to recreate Keller et al.'s grasp patterns turns up in many unrelated fields. Professional SCUBA diver gloves trade function for warmth in order to extend dive times. A mitten is warmest while a glove with individual fingers is the most functional. Professional SCUBA diver gloves are a compromise having the thumb and index fingers free and the middle, ring, & little fingers together. This configuration affords the diver the basic prehension patterns of the hand while at the same time keeping the bulk of the hand warm. A system of this kind (where the timing, or speed, of thumb is controlled) can be configured so a single "open" signal drives all digits (fingers & thumb) back to their start positions and two "close" signals, one for the index and MRL finger drives and a second for the thumb drive, control hand closure.

2.0 DESIGN METHODOLOGY

If one considers the role of a single DOF thumb operating along a plane in the 45° to 55° range [3,4,5], it becomes apparent that the final prehension pattern adopted by a hand can be determined by thumb position control i.e. if the thumb engages both the index finger and middle-ring-little (MRL) finger unit, then palmar/tri-digital pinch (three jaw chuck) results. If the thumb only engages the index finger, tip prehension results where as if the thumb closes on the side of the index finger, lateral prehension results. Power grasps result from digit shape and a wide width-of-opening. Thus it is possible through appropriate control of the thumb, to obtain different hand prehension patterns.

The primary consideration given to the design of the multifunctional hand was the ability to perform the three major grasp patterns described in Keller et al.'[1] whilst maximizing the system's robustness. This goal was approached on three fronts; minimize the number of parts, standardize the actuation drive across all joints and minimize the number of degrees of freedom.

2.1 DRIVE SYSTEM

It was decided to base the design around a drive developed for another project – the VA partial hand drive [2]. The same drive was used for every DOF. This drive system has already seen significant development work and uses commercial off the shelf (COTS) components. As shown in Figure 1, the drives are stacked one on top of each other in parallel where the top drive actuates the index finger and the middle drive actuates the combined middle-ring-little fingers using spur gears. The bottom drive unavoidably uses custom bevel gears to power the angled thumb. Each drive uses a Faulhaber 1724 DC motor, connected to commercially available multi-stage planetary gearing to give a 1023: 1 reduction ratio. With a 6 Volt input and 60% drive efficiency (90% per stage), the drive has an output velocity of 97.1 degrees per second with a stall torque of 16.3 Nm. Experimentally however, planetary gear failure was found to occur at 7 Nm. In terms of grasping, this drive is capable of producing 22.2lb_f (100 N) of pinch force; well within Piezer et al's [13] standard.

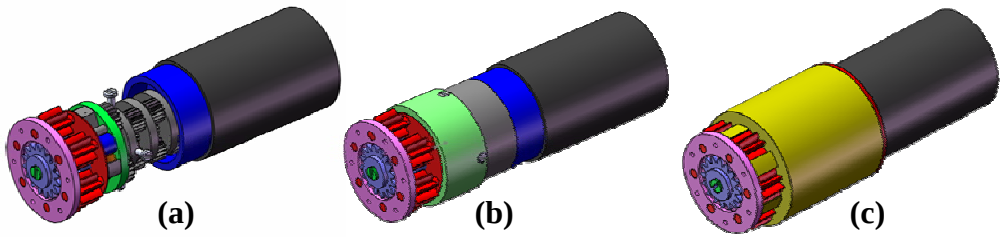

Figure 1 – Drive Train for the Multifunction Hand. The torque is generated from a five stage planetary gearing system with a backlock clutch before the output stage (a). The assembly is self-contained with two annulus' (b) while the the drive's output is the carrier of the last planetary stage (c).

2.2 FINGERS

Studying examples of less than clinically successful hand prostheses illustrate the necessity for simplicity in design [4,7,8,9,11]. In general they are complex devices where each finger contains several degrees of freedom. In contrast, the most clinically successful devices are single DOF designs, actuated at the metacarpophalangeal (MCP) joint that exclusively employ monocoque digits [12]. By employing a monocoque design and carefully choosing the proximal interphalangeal (PIP) and distal interphalangeal (DIP) angles of the index and middle fingers, in conjunction with the thumb's arc of contact, many grasp patterns were found to be attainable.

2.3 THUMB

Geometrically, the most challenging aspect of the design was obtaining a correct thumb angle that travels in a path where it can meet the index or middle fingers to form several grasp patterns. Work done by Lozac'h et al [4,5] stated that the thumb can attain near full functionality by actuating it at the carpometacarpal joint (CMC) at an angle between 45° and 55° relative to the to the top plane of the index finger. In addition to actuating the CMC however, Lozac'h also actuated the interphalangeal joint (IP) of his fingers. By emphasizing robustness in the digits, this design used a monocoque thumb design. A monocoque model however dictated that the design slightly deviate from the single 45°-55° CMC model and adopt a compound angle methodology as seen in Figure 2. This hypothesis was tested in CAD software and rapid prototypes. In order to keep the anthropomorphic nature of the design, the CMC thumb angle was increased to 60° with two 157° bends in series, ventral to the palm approximately one inch from the thumb's axis of rotation. The result of the ventral bends was that the thumb moves in a multiplane arc (Figure 2) that was able to meet the index and middle fingers at different specific locations to create the different grasp patterns.

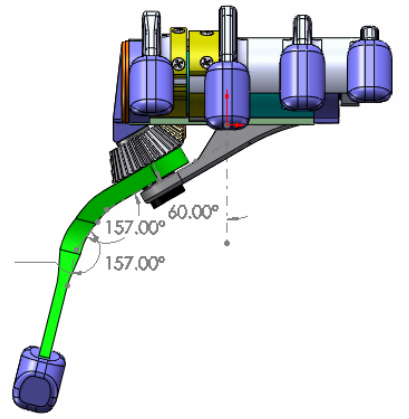

Figure 2 – Illustration of Compound Thumb Angle

3.0 DISCUSSION

The assembly was anthropomorphically designed to fit inside the volume of a 50th percentile female hand. Without the digits, the palm, which houses all three drives and its supporting structure, has the dimensions of 61mm wide, 73mm tall, and 24mm in depth. The base of the device was designed to interface with a yet to be designed wrist flexion unit. A first generation prototype (Figure 3) found that with careful selection of finger and thumb angles, several grasp patterns are attainable by controlling when the thumb comes into contact with the index finger. Thumb flexion after full index finger flexion yields lateral prehension (Figure 3a), equal flexion periods yield tip pinch (Figure 3b), while early thumb flexion relative the index and MRL fingers give palmar prehension (Figure 3c). During a grasp, the hand can anthropomorphically open to 5 inches, enough to pick up a ceramic jar or a can of pop/soda.

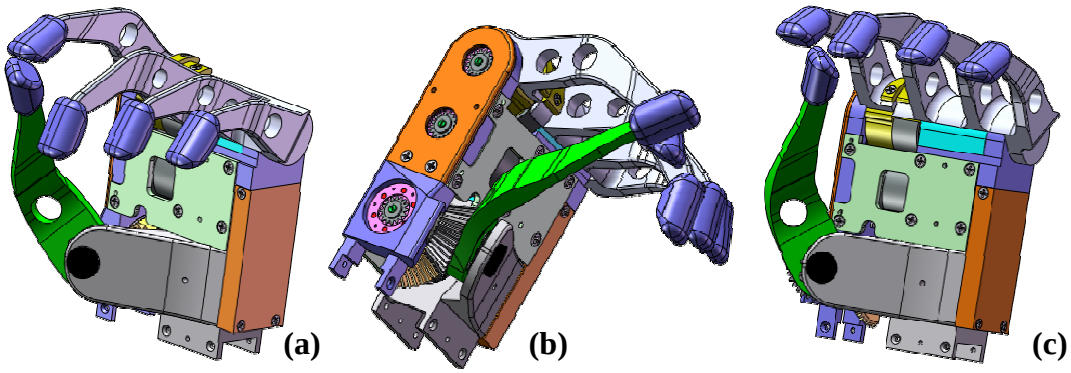

Figure 3 – Attainable Grasp Patterns for the Multifunction Hand with Three Degrees of Freedom

To control this hand, a three/four site EMG multifunction controller is under development. The key to controlling this device is accurate positioning of the thumb with respect to the index and middle fingers. This may be accomplished with a three surface electrode sites using end-state control for the whole device or proportional control for each DOF. The electronics are being developed at Sigenics Inc. and will be placed on the flat surface of palm's dorsal plate.

4.0 CONCLUSION

The design of new multifunctional hand prosthesis that we believe will be robust enough to ensure clinical success has been described. The hand has three common drives, one for each degree of freedom; the index finger, the combined middle-ring-little fingers and the thumb. Instead of increasing functionality and thereby decreasing robustness by further adding degrees of freedom, this design successfully positioned the thumb whose path of travel was common to the final thumb position of several grasp patterns. The result was that problems related to creating different grasp sets ceased from being a mechanical issue and primarily became a controls issue. Future work on this project aims to test the novelty of the design and have it tested with a subject.

5.0 ACKNOWLEDGEMENT

This work is supported by the Department of Veterans Affairs, Rehabilitation Research and Development Service, and is administered through the VA Chicago Health Care System, Chicago, Illinois.

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A NEW ACTIVE SHOULDER PROSTHESIS: FROM THE DESIGN TO THE FIRST CLINICAL APPLICATION

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ABSTRACT

INAIL and DIEM prototyped a new externally-powered prosthetic shoulder developed for interscapulothoracic and first-proximal trans-humeral amputees. The articulation consist of two connected powered joints that allow the elevation of the upper arm in any vertical plane passing through the shoulder centre of rotation. The development of the mechanism is the result of a rigorous approach, which made it possible to systematically combine both the technical and the clinical aspects involved in the design of a prosthetic device.

The prototype underwent laboratory tests needed to evaluate the mechanism's performance (e.g. the maximum payload) and the electrical requirements (e.g. the current draining). Based also on the results retrieved from these tests, an on-board embedded controller was implemented. The electronic unit can control a prosthesis with up to five powered joints and can manage different control strategies, according to the amputees' preferences.

The prototype with the embedded control system has been recently integrated within the prosthesis, provided with hand, wrist and elbow, of a proximal trans-humeral amputee who firstly tested the new device.

This paper provides an overview of the development of the actual prosthesis, reports the main patient's feedback and outlines the future developments.

1. INTRODUCTION

Nowadays no powered prosthetic shoulders are available on the market: current prosthetic solutions comprise at most passive articulations with locking mechanisms or frictional spherical joints. The orientation of these devices can be adjusted with the help of the sound limb, if present, or using fixed points in the surrounding environment, in the case of bilateral amputees. These solutions, due to the restricted mobility offered and the difficult passive adjustment required, preclude the possibility to perform important activities of the daily living, e.g. those requiring above-the-shoulder reaching.

Prototypes of actuated shoulder mechanisms have been proposed in the literature [1–4], but they do not appear to be easily applicable in the standard clinical practice.

In order to overcome these functional and technical limitations, the INAIL Prostheses Centre and the University of Bologna, developed a 2-degree-of-freedom active shoulder suitable for interscapulothoracic and first proximal trans-humeral amputees [5]. The new prototype is fully compatible with current commercial components for myoelectric prostheses. This paper provides an overview of the development of the prototype and reports the main feedback from a patient following a preliminary clinical assessment.

2. MATERIALS AND METHODS

2.1 Development of the shoulder prototype

The development of the prototype was based on a methodology which takes into consideration the patients' requirements and the need to integrate the final mechanism into the prosthetic arm already used by the patient (if any). In particular, the functional performance of the device was always balanced with the actual wearability of the resulting artificial arm (otherwise the risk was to design a high-performance device finally refused by amputees e.g.

because of a high weight, an intricate control or bad appearance). The methodology comprises different steps, briefly overviewed below.

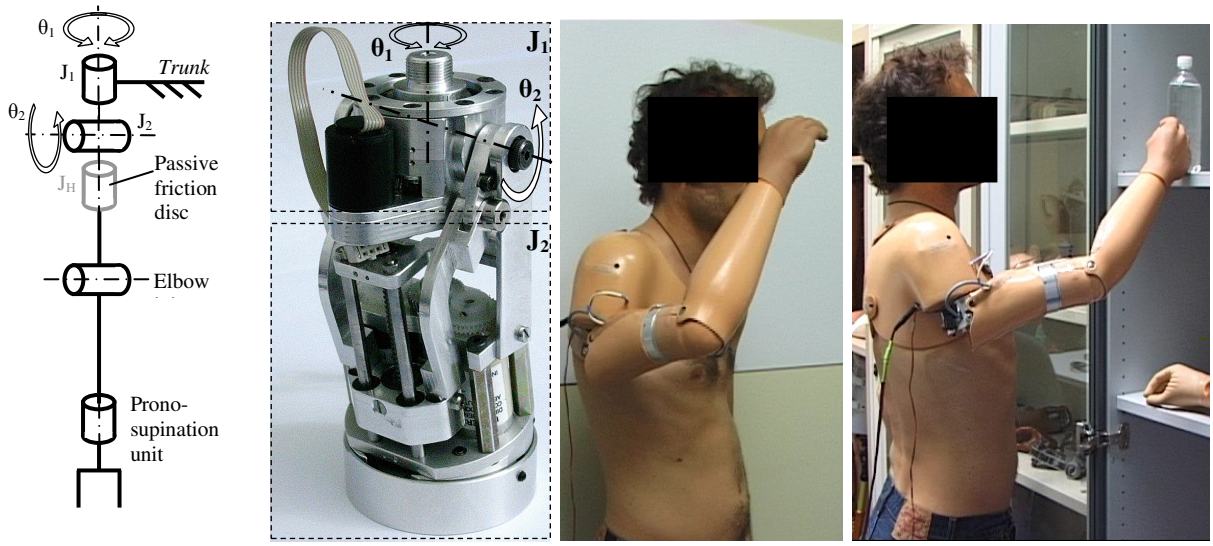
1) Kinematic Simulations – In order to determine the topology and geometry of the shoulder mechanism, a limited set of motor tasks were simulated by alternative kinematic models. Models differed in the combination and number of degrees-of-freedom (DoFs). The activities (e.g. feeding, personal hygiene, dressing and generic manipulation) were selected to be the most significant for the functional autonomy in the everyday living of an upper-limb amputee. The criteria used to select the final model were, among the others, the ability to execute the activities and the lowest kinematic requirements. From the results a shoulder model formed by two powered revolute joints with incident and orthogonal axes and a passive joint was selected. The two actuated DoFs allow a spherical motion of the arm, while the passive DoF allows the passive humeral intra-extra rotation [Fig 1(a)].

2) Feasibility – The selected shoulder model was implemented in a detailed mechanism with a CAD software. As from point 1), two connected powered joints, J_1 and J_2, and a passive one, J_H, form the articulation [Fig. 1(a)]: the active joints respectively actuate the rotation θ_1 about a vertical axis (fixed to the prosthesis socket) and the rotation θ_2 about a horizontal axis, whose orientation is determined by θ_1. In other words, the actuation of the first joint selects the vertical plane along which the second joint elevates the upper-arm. Joint J_1 is a simple actuated revolute joint, whereas J_2 is basically composed by an inverted slider-crank mechanism. This latter makes it possible to obtain a high torque without the use of a speed reducer with a great transmission ratio (and thus without low efficiency and high bulk). The passive revolute joint is a friction disc currently used at the INAIL Prostheses Centre and it is simply integrated at the extremity of the second joint.

3) Design and prototyping – Kinetostatic simulations of important activities of the daily living were carried out by a 3D-model of a full prosthetic arm including the shoulder mechanism. The retrieved data were used to dimension the actuators and the power transmission chains. After this stage, the mechanism was finally prototyped [Fig. 1(b)]. J_1 is driven by a commercial DC motor with nominal voltage $U_N = 6$ V, maximum output power $P_{max} = 4.55$ W, stall torque $M_H = 21.2$ N·mm. The total reduction ratio of the kinematic chain is $i_1 = 1{:}1050$, provided mostly by a commercial Harmonic Drive (GE) reducer. Similarly J_2 is driven by a commercial DC motor ($U_N = 12$ V, $P_{max} = 17.00$ W, $M_H = 80$ N·mm), whereas the total reduction ratio is not constant, depending on the parallel mechanisms configuration: the maximum value achieved is $i_2 = 1{:}1187$.

4) Bench tests – A series of tests were performed on a test bench specifically developed at the INAIL Prostheses Centre to experimentally characterize prosthetic devices. In particular the joints' maximum performance (in terms of velocity and payload) and their mechanical efficiency (varying with velocity and payload; average values $\eta_1 = 0.25$ and $\eta_2 = 0.45$ for J_1 and J_2, respectively) were estimated [6]. Moreover, several information related to the control were retrieved, and in particular about the current draining.

5) Controller design – An on-board control unit was designed and manufactured based mainly on the maximum current and voltage required by the motors, and on the need to maintain the compatibility with commercially available prosthetic components. The control unit consists of a Microchip PIC18F4431 microcontroller and 5 motor drivers. Receiving as inputs at most 4 EMG signals, the control unit can engage the 5 motors of the prosthetic arm, i.e. of shoulder, elbow, wrist and hand. By referring to the common control scheme where the joints are sequentially activated one at a time, 3 control strategies for joint switching were implemented in the controller: EMG co-contraction, 3rd electrode selection and double command.



(a) (b) (c)

Figure 1. (a) kinematic scheme of the prosthetic shoulder; (b) prototype of the mechanism; (c) an INAIL patient wearing the complete prosthesis equipped with the new prosthetic shoulder.

2.2 Clinical tests

A complete prosthesis, equipped with the new shoulder prototype and the control unit, was fitted on a patient for a one-day-test [Fig. 1(c)]. The patient was a male, 32 years-old, first proximal trans-humeral amputee with a very short stump at the right side and with a partial complex amputation of the left hand. The patient commonly uses a myoelectric prosthesis with hand, prono-supination and elbow joint in his daily life; two EMG electrodes are placed inside the socket. The subject controls the joints one at a time, cyclically switching between motors by means of a traction-switch engaged with an ante-retropulsion of the shoulder girdles. The control strategy was not modified with the introduction of the shoulder mechanism.

The patient used the new prosthesis for about 3 hours, executing common activities of the daily living which involved the use of the shoulder (in particular, but not limited to, the grasping of objects placed at heights above the shoulder-level, Fig 1(c)). To collect the feelings of the patient about the current prototype and for guiding in future developments, at the end of the test the patient completed a simple VAS questionnaire.

3. RESULTS

The questionnaire main outcomes are summarized in Table 1. The prosthesis weight (2.8 kg) is tolerable in the resting position (8/10) and well balanced (9/10) with respect to the sound limb. On the contrary, the load at the stump during flexion of the arm determines an excessive strain (3/10; above all when elevating the arm with the forearm completely extended); the velocity of the shoulder joints (about 40°/s) is good (8/10), whereas the payload (limited to 0.5 kg in order not to cause pain to the stump) should be rised to 1 kg (6/10); the noise is unacceptable (1/10); the control strategy is effective but not efficient, because too slow due to the high number of motors to be controlled (4/10); finally, its appearance is acceptable (7/10).

Question	**Patient's opinion** *[1 = not at all; 10 = completely agree]*
The weight of the prosthesis is tolerable	8
The prosthesis is symmetric with respect to the sound limb	9
The load at the stump is tolerable when using the prosthesis	3
The velocities of the shoulder joints are adequate	8
The payload at the hand is sufficient	6
The noise level of the shoulder joints is tolerable	1
The control strategy is simple	6
The appearance of the prosthesis is human-like	7
The donning and doffing is simple	9

Table 1: the most significant patient's opinions about the prosthesis equipped with the new shoulder mechanism.

4. DISCUSSION AND CONCLUSION

From a technical viewpoint, the prototype showed to be consistent with expectations, both considering range of motion and electromechanical performances.

The new complete prosthesis, thanks to its high mobility, can greatly improve the functional autonomy of patients with a high level of amputation.

However, the clinical tests revealed margins for improvements. Firstly, the sequential control strategy via traction-switch does not exploit the high mobility of the prosthesis. Alternative strategies for a more direct switching of the commands are needed. In particular, a voice control system integrated with the EMG-based control that allows the patient to directly select the joint to be activated seems promising [7]. Secondly, the noise of the mechanism will need to be reduced by reconsidering the quality of the manufacturing and/or design of some components (e.g. gears). Finally, in order to guarantee the proper wearability of the prosthesis, the socket should be re-conceived (with a wider contact area) to reduce the load on the patient's stump and thorax during the functioning of the arm.

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THE WILMER 2DOF-WRIST PROSTHESIS FOR TODDLERS

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ABSTRACT

The WILMER 2DOF-Wrist Prosthesis is designed to provide passive wrist flexion and passive pro- and supination is one compact unit. Therewith, the functional possibilities of a terminal device are increased while at the same time compensatory movements of trunk, shoulder, and arm are reduced. The WILMER 2DOF-Wrist provides 360˚ rotation and 120˚ flexion-extension. The wrist can be locked at 12˚ intervals for rotation and 30˚ intervals for flexion. The locks of the wrist rotation and the wrist flexion are simultaneously released by pressing two buttons located diametrically opposite of each other on the side of the wrist unit. In this way, the sound hand is able to support the terminal device and guide it to the new position desired. By releasing the buttons, the wrist automatically locks in the nearest locking position available. The mass of the prototype is 22.3 g.

INTRODUCTION

Upper extremity prostheses usually comprise a terminal device attached to the prosthesis by a wrist unit that provides passive pronation and supination. Clinical observation of users of arm prostheses revealed that performing bi-manual tasks at the midline of the body requires compensatory movements in the shoulder due to the lack of (passive) wrist movements, especially wrist flexion. Perhaps for this reason many users of prosthetic devices put wrist flexion-extension high on their wish list [1]. Existing wrist flexion prostheses often rely on friction to maintain the position chosen [2]. This always implies a compromise between the effort needed to change the wrist flexion angle and the capabilities of the wrist to sustain loading. Furthermore, many existing wrist flexion prostheses are designed to be added on top of the wrist unit that provides pronation and supination [2]. Recently, many attempts to achieve a wrist flexion possibility in addition to pronation-supination have been reported. Some of these efforts aim at a passive solution [3, 4]; others aim at an external powered solution [5, 6, 7]. All developments are aimed at devices to be used by adults. Therefore, it was decided to try and develop a passive wrist prosthesis for small children that offers pro- and supination as well as wrist flexion with a positive locking of the position chosen.

METHOD

For the design of the new wrist prosthesis a 4-year-old child with a unilateral arm defect was chosen as a reference. The overall dimensions were set at a maximum of Ø 35 x 15 mm. The wrist should provide wrist flexion in at least three positions of -60°, 0°, and 60°; as well as wrist rotation up to 90° pronation and 90° supination, thus covering most of the average human range of motion [8, 9]. The minimal requirements

for the load were set at 15 Nm for the wrist rotation torque; and 20 Nm for the wrist flexion torque, based upon the idea to be able to support the body weight on the terminal device. The overall mass of the wrist prosthesis should be as low as possible, and preferably not exceed 20 g.

RESULTS

A new wrist prosthesis providing passive wrist rotation as well as passive wrist flexion has been designed, constructed, and build, Figure 1. The overall dimensions are Ø 35 x 15 mm.

Figure 1. The new WILMER 2-DOF-wrist prosthesis for toddlers. It provides passive wrist flexion and passive wrist rotation.

The wrist provides 5 flexion positions at -60°, -30°, 0°, 30°, and 60°, Figure 2, and 360° of wrist rotation with 30 locking positions at 12° intervals.

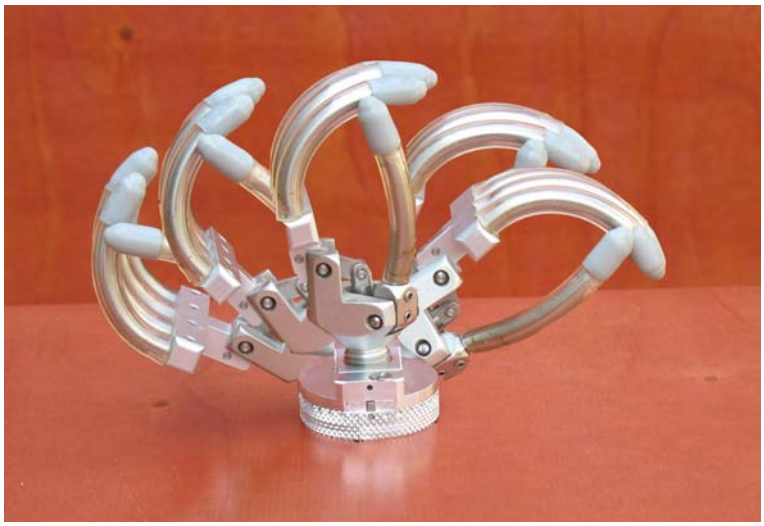

Figure 2. The new wrist provides 120˚ flexion-extension in 30˚ intervals.

The locks of the wrist rotation and the wrist flexion are simultaneously released by pressing two buttons located diametrically opposite of each other on the side of the wrist unit, Figure 3. In this way, the sound hand is able to support the terminal device and guide it to the new position required. By releasing the buttons, the wrist automatically locks in the nearest locking position available.

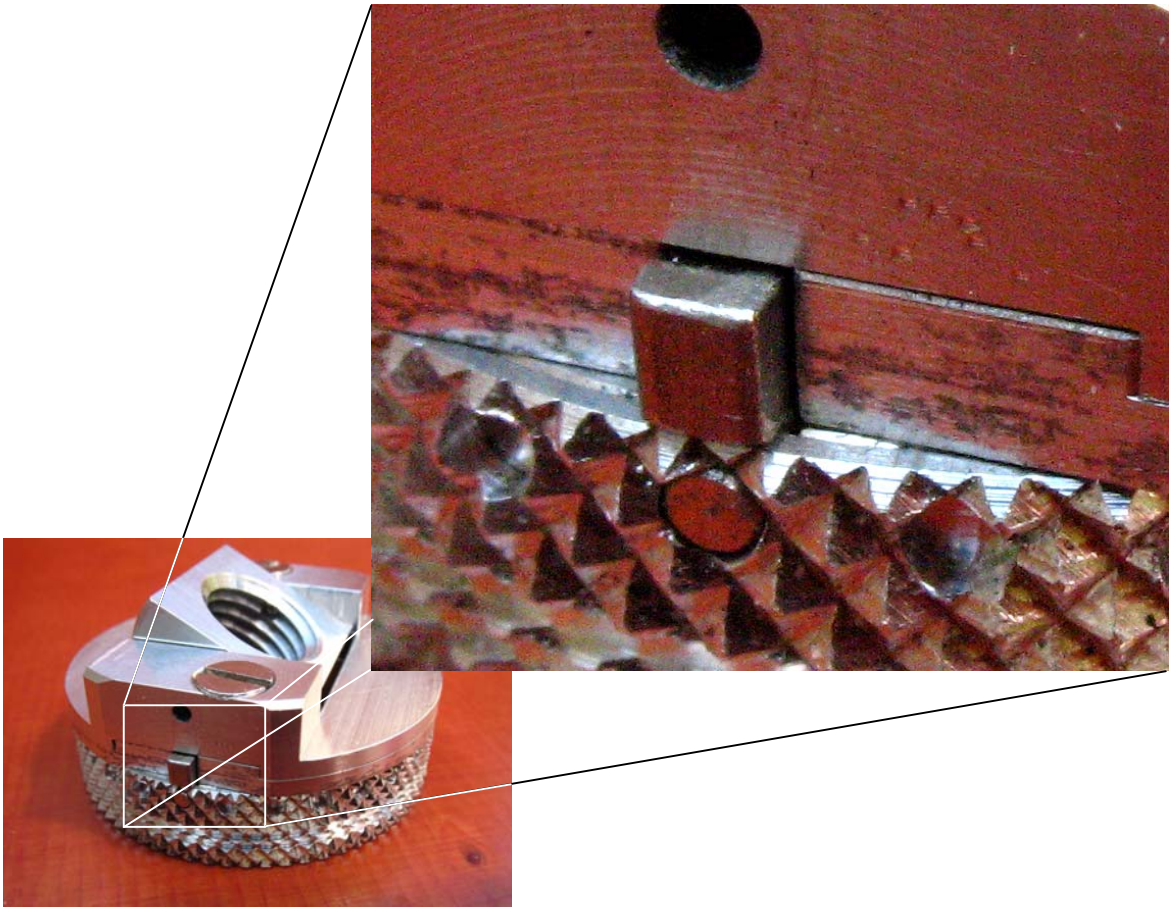

Figure 3. By pressing two buttons, located diametrically opposite of each other, the wrist rotation and the wrist flexion are simultaneously unlocked.

The maximum allowable loads are calculated at 24 Nm for the wrist rotation torque, and 21 Nm for the wrist flexion torque. The mass of the prototype is 22.3 g.

CONCLUDING REMARKS

A new two degrees of freedom wrist prosthesis has been designed and build. The technical specifications of the prototype exceed most of the requirements set; only the mass is slightly higher. Moreover, the new wrist design compares favorably with existing commercially available wrist units: the WILMER 2-DOF wrist combines wrist flexion and rotation into one unit, has a mass (much) lower than most of the other wrist units, while it sustains comparable permissible loads. For the future, clinical trials should

endorse the functional benefits of the new wrist. Subsequently, extending the 2-DOF-wrist design into a series of different sizes to accommodate different age groups and/or different terminal device designs is intended.

ACKNOWLEDGEMENT

J.M. Nicolai contributed significantly to the design of the WILMER 2-DOF wrist prosthesis as part of the requirements for the award of the Bachelor's degree in Mechanical Engineering at Rijswijk University of Professional Education, the Netherlands. We thank our clinical partners of the rehabilitation centers De Hoogstraat and Sint Maartenskliniek for their cooperation.

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EXPERIMENTAL CONSIDERATION ON THE FACTORS WHICH CAUSES VARIATION IN FITTING SURFACE EMG INTERFACE

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INTRODUCTION

Electricalmyography or myoelectric signal is used to voluntarily control current commercial electric prosthetic modules in the upper limb prosthetics. One of the technological questions that are rarely discussed is the assessment of fitting or tuning the EMG sensors to gain suitable and robust control signal. Such methods are mostly neglected by laboratory researchers because workable signals can be obtained through trial and error for short-period experiments, and for clinical use, humans are more adaptable then machines. However, the reliability of the interface is a key feature for promising everyday use of a device, and therefore, an assessment strategy and a design method for assembling a durable myoelectrically controlled prosthetic arm is expected. Furthermore, quantitative data is advisable for product design which requires robustness in long life use.

In this paper, we applied a Quality Engineering technique [1] to investigate the factors in installing EMG sensors for generating activation (ON/OFF) control signal. Eight influential factors on fitting surface EMG electrodes for prosthetic hand control were selected based on heuristics and De Luca's relation diagram [2], and a multifactor experiment was conducted as a pilot test on a single able-bodied subject.

METHODS

We applied an evaluation process of robust engineering, which is an experimental design method in the Quality Engineering, to assess the relationship of the factors that destabilize the EMG controller response. In this experiment, the main function of the EMG controller is defined as the response efficiency between the intended joint movement frequency and the controller's output frequency, in which the ideal case is 1. As for the factors that influence the function, which are named control factors in QE, eight items were selected for the investigation: electrode contract pressure, electrode displacements toward long- and short- axes, sensor orientation, passband- and cutoff- frequencies of the low pass filter, window size of the enveloping process, and activation threshold. See Table 1 for description of the

control factors and their levels. Furthermore, three items were selected as noise factors. The noise factors determined in QE are items which cause variance of the system's performance, but are clinically non-controllable. In this case, 1) with and without moisture on the skin at the electrode, 2) Heavy and light muscle fatigue, and 3) arm at body-side and front reach position, are selected as the 2-levels of the noise factors.

Three-pole dry EMG sensors with differential amplifier (Personal-EMG, Oisaka Electronic Device Ltd., Japan) were used as test instrument in the experiment, (IEMG Time Constant: 66 msec, CMR: 104 dB). This system is capable of simultaneous recording and monitoring two EMG channels (Max. 8) with band eliminating filter of 60 Hz, power frequency of West Japan. The myoelectric signals were sampled at 3 kHz with 12-bit AD converter and recorded. Then, the signals were processed offline with an IIR filter, enveloping algorithm, and discriminant function with activation theshhold.

Before starting the experiment, Informed consent was obtained from the subject: a 22 year-of-age male with no-experience of EMG control in advance. His left arm, which is his non-dominant side, was used as the signal source. The EMG sensors were place on the surface above the extensor digitorum and flexor digitorum superficialis and fixed with an adhesive tape. During the trial, movement cycles of 45, 60, 75, 90 ppm were presented as a beep sound from a digital metronome, while the EMG wave signals were visually feedback to the subject through a monitor. The trials were conducted with 15 second recording with minimum of 1-minute intervals. The time keeper count down the start and the subject was to continue a 'flexion–reposition-extension-reposition' movement cycle as in Figure 1. Action potentials were confirmed as distinct signals during the 15-minute rehearsal of each movement cycles. The EMG signals of the Maximum Voluntary Contraction, were recorded before and after the test for each joint movement cycles.

Table 1. Control factors and level setting of the EMG control function

		Level 1	Level 2	Level 3
A	Electrode contract pressure	Firm	Mild	
B	Short-axis displacement	0 mm	3 mm	5mm
C	Long-axis displacement	0 mm	5 mm	10 mm
D	Sensor orientation	0	+5 deg.	+10 deg.
E	LPF passband frequency	900	1000	1100
F	LPF cutoff frequency	+10%	+15%	+20%
G	Enveloping window size	10	20	30
H	Activation threshold	8% MVC	10% MVC	12%MVC

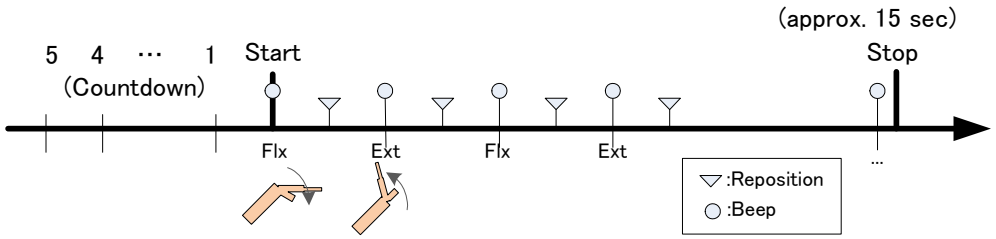

Fig.1. Diagram of the Protocols for EMG recording trials in time series

RESULTS

With the configuration of 8 control factors as in Table 1, experimental conditions of 18 combinations of the control factors levels were generated by using a technique in experimental design method. The conditions were applied to the electrode on the Flexor muscle, while the extensor side remained unchanged. A mixture factors with 4 levels were created from the noise factor to reduce the experiment number. Consequently, with 4-level noise factors, 4-level input, and 18 combination of control factor, total of 288 experiments were conducted. The maximum activation cycle time is computed from each result, and converted to frequency. SN ratio η, which is the reciprocal measurement of error variance, and sensitivity S, which is the coefficient value of the input-output relation, is computed based on defined functions for QE dynamic characteristics analysis. Diagrams in Figure 2 and 3 are obtained as the computed analysis result.

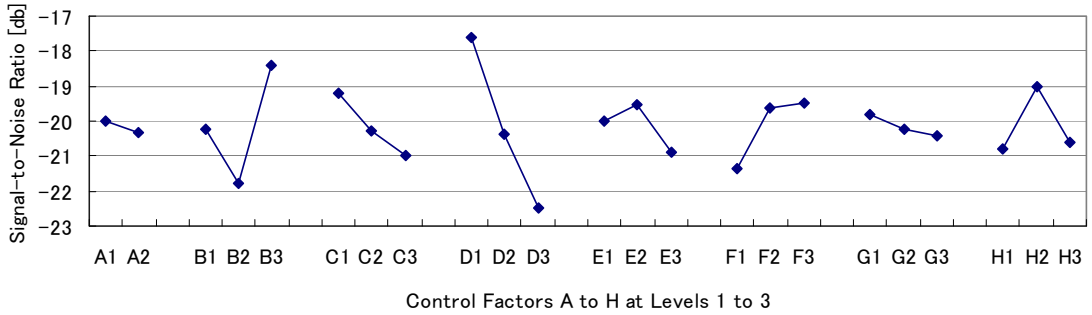

Fig. 2. SN ratio of the EMG controller function

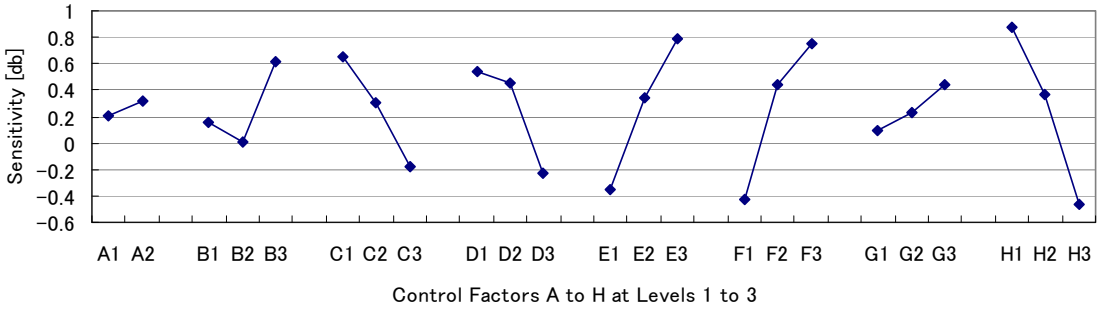

Fig. 3. Sensitivity of the EMG controller function

DISCUSSION

From Figure 2, the three levels in D and B, which are the sensor orientation and sensor short-axis displacement, shows the largest difference among the levels. This results show that sensor oriented in the muscle fiber direction is most effective on stabilizing the response of the switching frequency. Moreover, the sharp declines in the sensor orientation results explains that precision is required or installation. The result of short-axis displacement illustrates that precision is require for locating the sensor, however the “V” curve is unexplainable at the moment. Possibility of the cause may be the electrode dimension, relation to the innervation zone, or other physiological causes. As for other control factors, the range of SN ratio are similar which means that there is no major order in setting these factors. The effect of A and G, which are the contract pressure and the widow size of the enveloping process has minimum influence to the function, and therefore, can be replaced by other prime factors or even omitted in later experiments. When taken together, a suitable condition of control factors can be chosen from the smallest SN ratio when the variance of the SN ratio is large, and for small SN ratio variance, the condition can be determined based on other criteria, such as sensitivity.

The result of Figure 3 shows that all control factors’ sensitivities tend to cross or remain near zero-decibel. This means that our functional model of the input-output relation is effectual and can be adjusted to a suitable condition. When comparing the result among the factors, variance of E, F, and H are larger then the other. This describes that LPF design and the threshold value has strong effect on dissipating the response characteristic, therefore requires caution is selecting the parameter.

To complete the experiment process of robust engineering, verification experiments are required to confirm the analyzed results and to prove that the selected suitable combination of the control factors’ conditions are capable of improving the robustness of the function. These experiments are planned in the near future. In addition, further testing on multiple subjects and regional differentiation will be our next target.

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ADVANCED SIGNAL PROCESSING TECHNIQUES APPLIED TO CROSS-TALK REDUCTION IN FOREARM S-EMG

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ABSTRACT

In spite of the great advances in the mechanical and electronic components of prosthetic hands, they still lack the high number of degrees of freedom present in the real human hand. That is due, not to technical deficiencies, but to the much reduced amount of independent control signals available when using surface electromyography (s-EMG) from the forearm stump or other artificial sensors. Cross-talk between adjacent muscles produces interferences that bury the s-EMG of the target muscle and reduce selectivity.

In a single case study, surface-EMG signals from an able-bodied subject's forearm were recorded with a surface, 5x13-electrode array while the subject performed eleven different isometric contractions. In order to reduce the cross-talk between s-EMG signals from different muscles, we applied a blind source separation (BSS) technique called JADE.

Although the results are not fully conclusive, they indicate that BSS techniques could provide an important reduction in s-EMG cross-talk and hence BSS is able to increase the selectivity of recordings for myoelectric control.

INTRODUCTION

Current myoelectric prosthetic hands have generally only one or two degrees of freedom (DOF), which allow only a very limited variety of movements. Moreover, their control is just in on-off or simple proportional mode. Therefore, users cannot yet control them in a natural way. This is possibly the main reason why most users do not choose to rely on myoelectrically controlled hands in spite of their high technical level.

On the other hand, robotic hands and arms have several DOF; however, they are not suitable as prosthetic limbs mainly because they need too many signals for their control.

Trying to combine the best characteristics of robotic and prosthetic hands, several Laboratories have developed new types of artificial hands with high (even above 10) DOF. To make those devices controllable in a natural way (for example, by agonist-antagonist pairs of muscles), many more than the usual two independent myoelectric signals should be collected from the muscles located in the stump of the user. In the forearm there are 19 muscles, all of them of small size, very close to one another, and often crossing and overlapping. Therefore, when trying to record surface electromyographic (s-EMG) signals, we have to face a very high level of cross-talk among the different muscles.

In the present work, we tried to solve the problem of cross-talk between forearm muscles by employing a blind source separation (BSS); the Joint Approximative Diagonalization of Eigenmatrices (JADE) [1].

We had already applied JADE for the separation of s-EMG signals into its constituent elements (motor units action

potential trains) [2]. In addition, other researchers had satisfactory applied other BSS technique for the reduction of cross-talk in forearm s-EMG signals [3]. However, the technique they used is based also on information gathered from the frequency domain, which might render it unsuitable for long-lasting recordings, as the frequency characteristics of s-EMG change with factors such time and fatigue [4].

METHODS

Data Acquisition

A healthy, male subject participated in this experiment after giving informed consent. He sat comfortable on a chair and his right arm was fixed to a mechanical device that blocked any movement of the hand and wrist (see Fig. 1), and measured the torque exerted by each of the isometric contractions, which consisted of: flexion of the distal phalange of digits 2 to 5 (from the index to the little fingers) and then of their medial phalange; after that, wrist abduction and adduction; and finally, wrist (forearm) pronation. Each contraction was performed three times, following a symmetric, 10-second, ascend & descent ramp with peak at 50% of subject's maximum voluntary contraction (MVC); each iteration was separated by a 3-second relax period.

A s-EMG recording was made for each of the isometric contractions with a 13x5 electrode array (model *ELSCH064* from *OT Bioelettronica*, Torino, Italy) placed on the anterior part of the forearm as shown in Fig. 2 in monopolar mode (far reference electrode –*ARBO* pediatric ECG- placed on the wrist). Each of its holes features a metal ring that was filled with conductive gel; the inter-electrode distance was 8mm. The signals were acquired at 2048 samples/s.

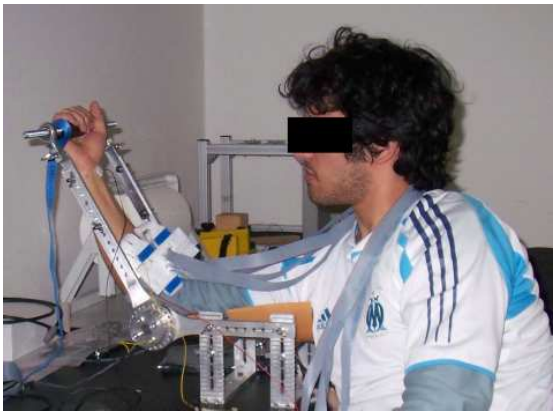

Figure 1. Experimental setup.

Signal Processing

Firstly, each of the 65-channel monopolar recordings was transformed into a bipolar 60-channel signal by column subtraction (direction perpendicular to muscle fibers).

JADE was then applied to the 60-channel for each of the 11 different contractions in order to obtain one representative channel for each of the contractions. The separation matrix obtained (the inverse matrix resulting from the mixing process), was then used, after normalizing, to create a weighting matrix that represented the best linear combination of channels for each of the contraction "fingerprints". Figure 3 shows a bi-dimensional representation of the obtained weight matrices corresponding to each contraction. An "importance"

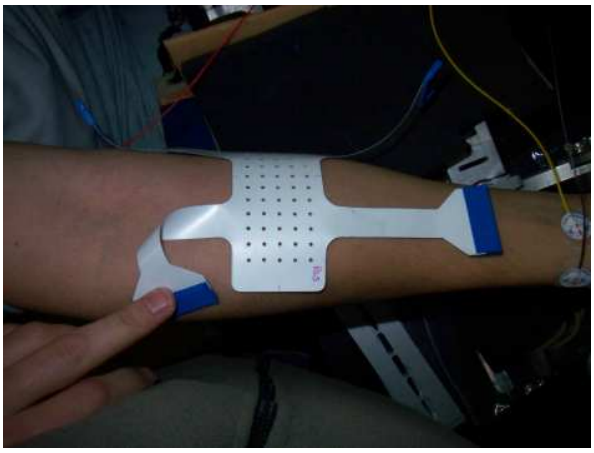

Figure 2. Placement of the electrode array

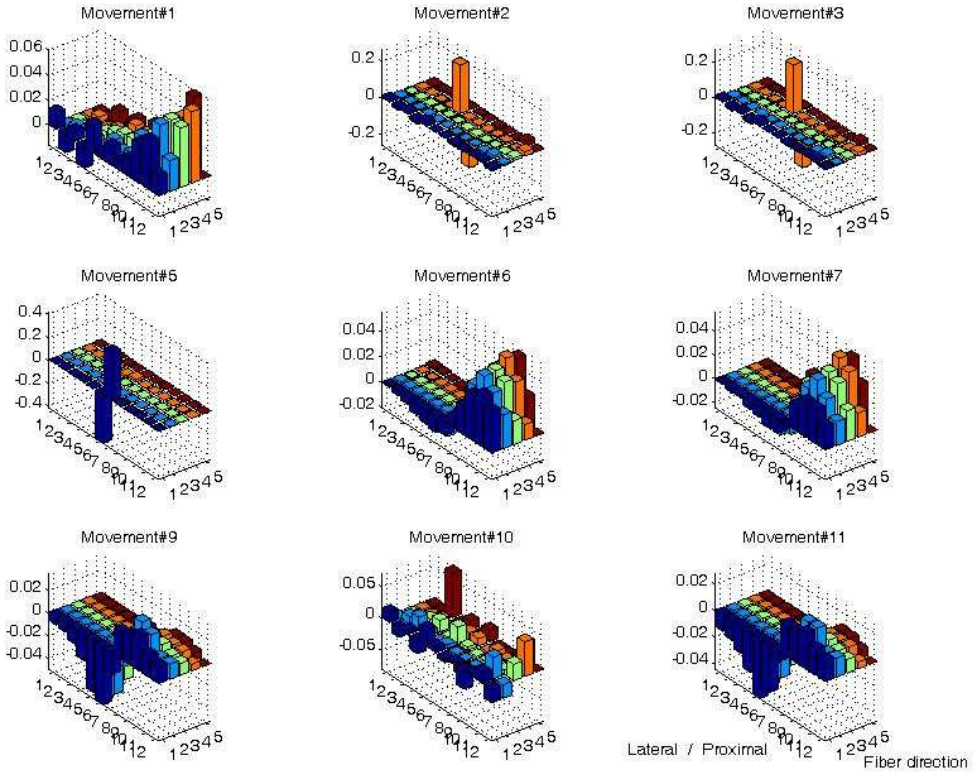

Figure 3. Weights for s-EMG channels for each of the contractions.

matrix was then created using for each of its lines the 1x60 matrix given by JADE.

This matrix was applied to each of the recordings. The most powerful channel of the output (in the sense of higher root mean square value), was then considered to be the representative contraction and therefore the one with the most powerful activity.

RESULTS AND DISCUSSION

Figure 4 shows the identified contractions. Out of the 11 contractions, six were correctly classified. Four of the remainder ones were just up to 10% less powerful than the winning contraction, and one of them was as powerful as the recognized contraction. It is worth noting that the entire processing was automatic, and no corrections were made when, for example, JADE algorithm did not yielded a representative s-EMG activity for the target contraction. For example, by adding some further signal processing such as filtering, better results could be obtained.

CONCLUSION

This approach represents a promising solution for diminishing the high level of cross-talk usually seen in s-EMG signals. It is also a faster and more economic way of classifying muscular activity form their s-EMG with respect to other sophisticated statistical or artificial neural network algorithms. With additional processing these results could be further improved.

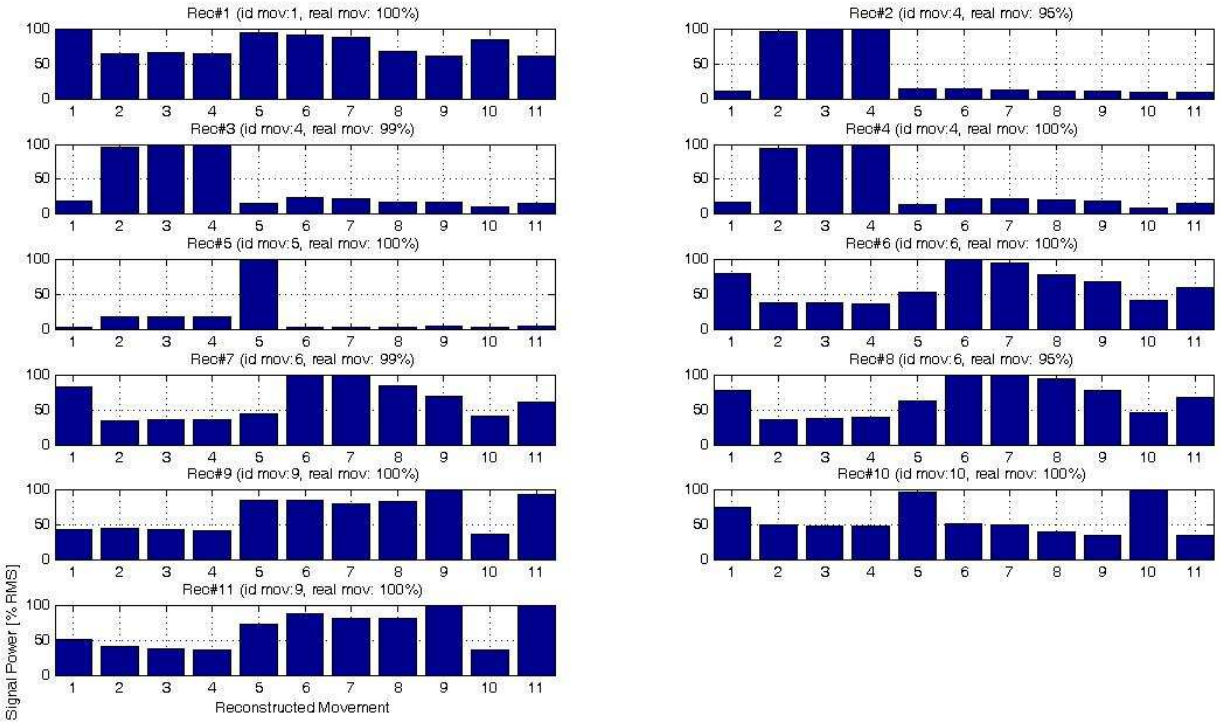

Figure 4. Results of the contractions identification.

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ACKNOWLEDGEMENT

The experiments were carried out at Prof. Merletti's LISiN Lab (Torino, Italy). The authors thank him and his group, especially to Dr. Introzzi, Dr. Vieira, and to Dr. Mesin, for their invaluable help and input.

A COMPENSATING SHOULDER JOINT TO ASSIST THE SHORT TRANSHUMERAL AMPUTEE

T. Walley Williams, III, MA
Liberating Technologies, Inc

THE PROBLEM TO BE SOLVED

The typical problem patient has a transhumeral amputation 30 to 60 mm below the axilla. No matter how well the socket is made, the moment arm in the socket is too little to sustain the four foot-pound torque of a typical advanced prosthesis when working out in front. Matters get worse if the prosthetic terminal device carries a load. A prosthesis that solves this problem must compensate for the gravitational torque generated by the prosthesis itself so that all user generated torque can be used to position the prosthesis and to support the terminal device load.

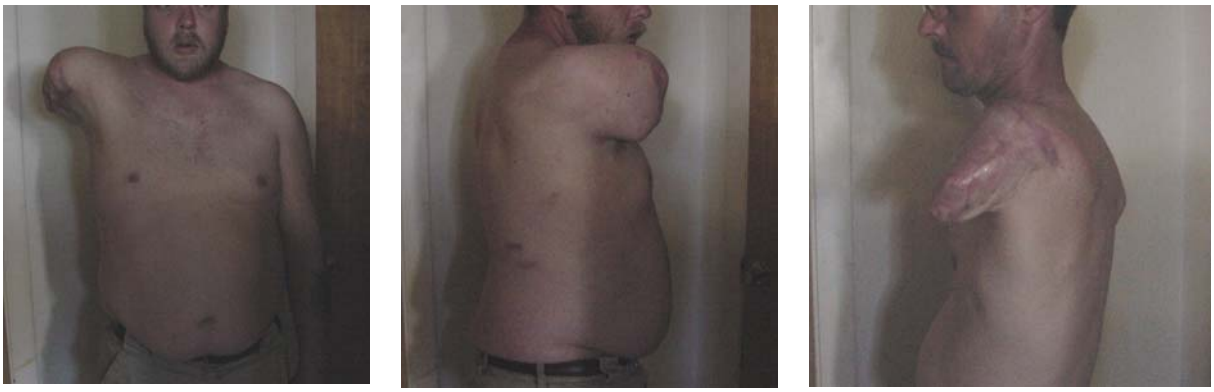

Figure1. Two patients with a common problem — a limb too short to support a distal load

Identifying the Problem

The problem became obvious when attempting to fit the two short transhumeral amputees shown above. While the amputee on the right has been lost to follow up, the one on the left has received several advanced sockets from Next Step in Manchester NH. He has a full range of motion without his prosthesis, but even with improved interface designs, he still has difficulty holding the prosthesis out in front at an angle sufficient to do useful work. As a manual worker, this is precisely what he needs. The compensation mechanism discussed in this paper has evolved since meeting this patient in 2006.

ONE INTERFACE FOR STABILITY AND ONE FOR CONTROL AND POSITIONING

The user's stabilization and suspension interface is a modification of the X-frame used for most shoulder disarticulations. The center of the X is moved down to accommodate the user's remaining arm making the frame look like a giant U-shape with the contacting arms wrapping around front and back. The four contact surfaces are located for maximum stability while still permitting the user full motion of the shoulder and arm.

The user's primary control interface is a short socket without the long stabilizing wings present on most transhumeral sockets. The exact shape of the new socket must be optimized for maximum production of flexion-extension and abduction-adduction torques. With a fleshy limb, a uniform cylinder couples these motions to the prosthesis only poorly. Randall Alley, CP and Matthew Albuquerque, CPO have made several sockets of advanced design for such patients to improved coupling between the humerus and the socket during motion. These advanced shapes without the stabilizing wings are ideal for a prosthesis with compensation.

The Two Sockets Are Joined by a Three-Element Two-Axis Bearing Mechanism

The first element is a thin but large-diameter flexion-extension bearing attaching to the humeral socket on the lateral side as close to the side of the arm as possible. To enable the amputee to flex and extend naturally, the axis of rotation of this joint must pass through the center of the humeral head. In the model in figure 2, the flexion-extension bearing is shown on the right with the strut that joins the humeral control socket below it. To fulfill the closeness constraint, the joint must be as thin as possible. The one discussed here is only 11mm thick. And it must permit the user to extend the arm 15° rearward and to forward flex 175°. The user's anatomy makes it probable that the plane of the flexion joint will be rotated forward of the saggital plane by 20° (See figure 5).

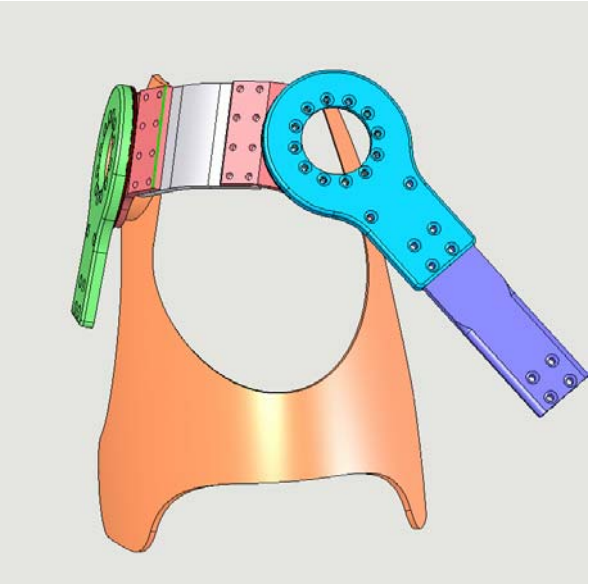

Figure 2. The bearing assembly elements are two flat, thin bearings and the cross-strut that joins them. This configuration is for a right shoulder. The left bearing is affixed to the frame and is the abduction joint, while the right bearing is for flexion.

The second element is a wide flat strut bent to follow the curvature of the user's shoulder. It mounts both the flexion and abduction bearings. As long as this strut is subjected to simple bends without any twist, the axes of rotation of the joints will pass through a common center. (See figure 4.) This strut must be adjusted to fit the user so that the common center also passes through the center of the humeral head. The angle between the two axes will typically be 80 ° to 90 °.

The third element is the abduction bearing on the left. Typically it will be similar to the flexion bearing. The fixed part of the bearing attaches to the X-frame, while the movable part attaches to the strut. The constraints on the abduction joint are that it be as close to the user's back as possible and that the axis of rotation must pass through the center of the humeral head.

DESIGN OF THE BEARINGS TO ACCOMMODATE REMOTE COMPENSATION

Since there is no practical way to compensate for large gravitational torques in a thin bearing assembly, the compensation forces must be applied to one element of the bearing by a remote mechanism. Solid linkages can do the job, but a cable is simpler. Consider the image in figure 3. A ring with a pulley groove and the plate above it trap the outer race of the main ball bearing. For the flexion joint, this plate is part of the strut joining the two bearings. The inner race and

Figure 3. The flexion bearing showing a cable in the pulley groove of the large-diameter outer race. An idler pulley redirects the cable to pass down the thin strut on the right into the arm. This strut attaches to the compensator below the remaining limb.

the remaining elements are attached to the humeral socket by the arm on the right. The all-important compensation cable passes around the large pulley groove until it is redirected by the small pulley associated with the second ball bearing. The other end of the cable is attached to the compensation mechanism in the arm distal to the user's remaining limb. While figures 2 and 4 show an adduction bearing identical to the flexion bearing, the final design will offer a free swing joint without the "handle". It should be apparent that there is little room on the back of the X-frame for a compensation mechanism.

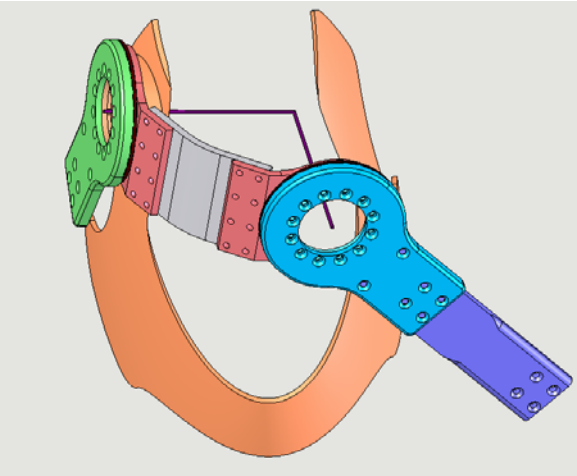

Figure 4. The two bearings are joined so that the axes cross. The cross point must also pass through the center of the humeral head

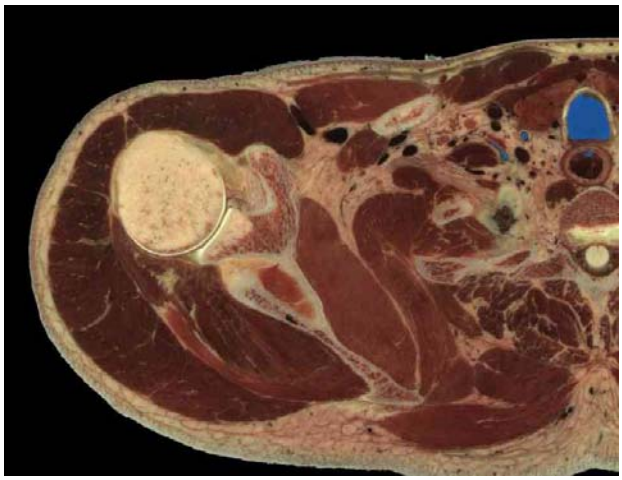

Figure 5. The glenohumeral head of a left arm is viewed in cross section from above. The joint location favors flexion about an axis that is tilted forward about 30°.

THE COMPENSATION MECHANISM

There are a number of constraints on the size and shape of the compensation mechanism. In the flexion case, it must be located distal to the user's residual limb or be sufficiently thin to fit along the lateral wall of the control interface between the inner and outer walls. Likewise, the adjustable-length strut that attaches the bearing to the compensator must be as thin as possible, and the mechanism should be as small as possible. Finally the compensation itself must vary with the angle of flexion with no compensation at full extension and a maximum at 90° of flexion. The maximum torque should be adjustable to accommodate users with varying arm lengths and types of prosthetic components.

Prototype LTI Compensation Mechanism

The bearings and compensators were developed together. Once a cable had been selected to link the two mechanisms together, it was obvious that the compensator should have a pulley the same diameter as the pulley in the bearing assembly. The bearing pulley requires a tangential force that is zero at full flexion and maximum at 90°. This then becomes the requirement for the compensator pulley. The ideal tangential force follows the sine function, and there are a number of classic mechanisms that can approximate this force. Figure 6 shows the prototype compensator which uses a gas strut to generate the required force. Gas struts are widely used to compensate for the weight of automobile hatch backs. To generate an ideal force, the gas strut should be long compared to the diameter of the pulley. In a prosthetic application an approximation of the sine function is good enough. The strut in figures 6-8 is short which causes

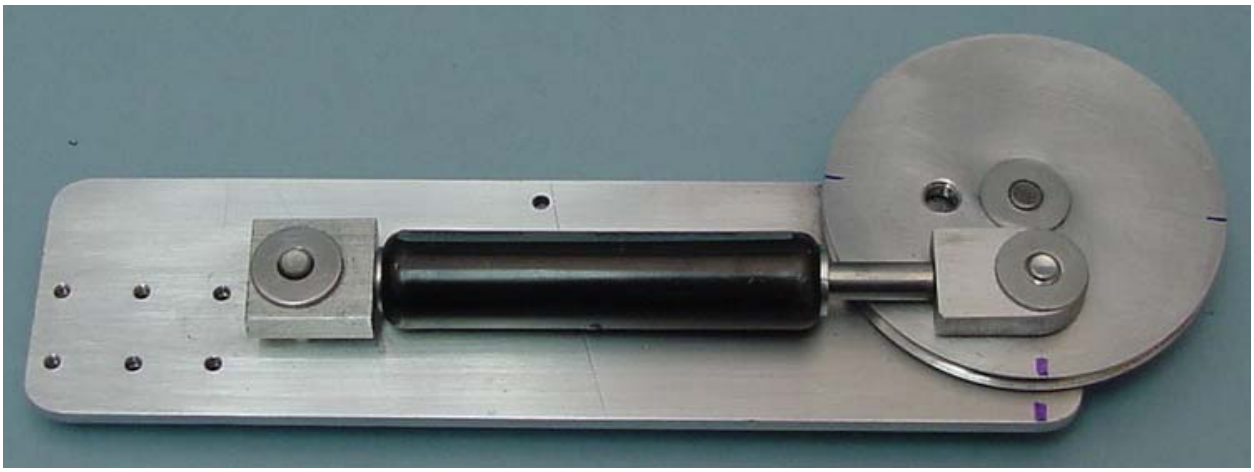

Figure 6. The gas strut is shown in the position of maximum tangential force. The cable from the bearing parallels the long axis of the assembly. The two marks show the location of the maximum force position of the pulley. Note that two thinner marks for zero compensation are on a line about 15° from line between the pulley and strut axes.

maximum compensation to occur at 105° rather than 90°. This is not a problem since the user can still generate considerable force with the short remaining limb. An important variable is the maximum torque. The gas struts are available in three force levels 200, 300, and 400N. By attaching the output of the strut to the pulley at three different radii, a total of nine equally divided force levels can be generated.

Problems with the Prototype

The prototype in figure 6 has worked for testing the concepts behind the design, but there has been a practical problem — the 400N force from the maximum force strut put too much cross torque on the pulley bearing. A redesign with a large diameter ball race will solve that problem. The other bearings will require attention too. Finally a thin molded cover will make it easier for technicians to mount the assembly in the arm.

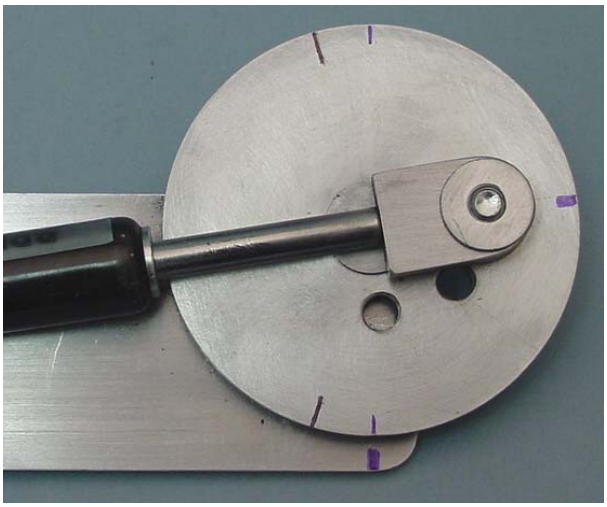

Figure 7. The fully extended strut produces no tangential force. The angle between this position and that of full force is about 105° not 90°.

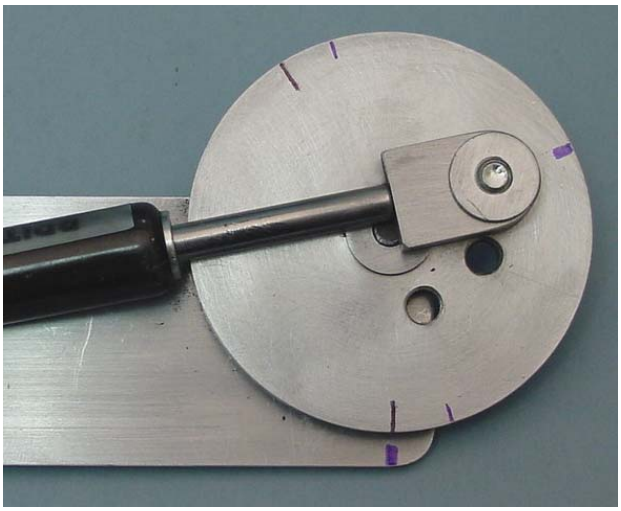

Figure 8. An extra 15° before hitting a limit stop facilitates free swing and tasks requiring extension. Note the two extra holes for the end of the strut.

EXPERIENCE FITTING PARTIAL HAND PROSTHESES USING PRODIGITS

Jack E. Uellendahl, CPO
Elaine N. Uellendahl, CP
Hanger Prosthetics and Orthotics

INTRODUCTION

Partial hand amputation proximal to the MCP joints are challenging to fit due to the limited space available for prosthetic mechanisms. Historically this group of amputees has been underserved by prosthetists. With the introduction of individually powered fingers, ProDigits by Touch Bionics, exciting possibilities for fitting externally powered finger prostheses are now feasible. The author has gained experience designing and fitting prostheses using ProDigits for persons with all fingers missing as well as with the thumb remaining.

PROSTHESIS DESIGN CONSIDERATIONS

When the thumb is missing, but space does not allow for a mechanical thumb, a rigid post has been employed. Alignment of the thumb post is critical. When the thumb is positioned well several prehension patterns can be approximated allowing grasp of objects ranging from a pop can to tip prehension of objects as small as ½ inch. When the thumb metacarpal is intact and mobile all attempts have been made to utilize this motion actively (Figures 6-8). A silicone thumb extension is intimately fitted to the bony anatomy to harness as much motion as possible. When this is possible, the user has direct control of thumb position and may benefit from direct force feedback through the prosthetic extension to the intact skin. The author feels this feedback is especially important when fitting persons with bilateral upper limb amputations where it is likely that the partial hand will serve as the dominant hand and the ability to sense the force applied provides the user with important information (Figure 8). In cases where the thumb is replaced with a mechanical passively positioned thumb, two designs have been used. The first is a two position thumb allowing for a narrow or wide opening (Figures 4,5,8,9). This thumb has been adapted from the APRL mechanical hand manufactured by Hosmer. The second is a prototype thumb with friction control of flexion/extension and rotation designed by Matt Mikoz, CP (Figures 1,2,11). Attempts to fit the ProDigit thumb have been unsuccessful to date due to space limitations. The potential advantages of the powered thumb are increased grip strength and provision of a wide variety of opening widths. The disadvantages are the need for space, both for the thumb mechanism as well as a second VariGrip controller dedicated to its function. A prototype thumb rotation bracket has been developed but is not yet in clinical use (Figure 3).

Custom silicone interfaces have been used to provide excellent suspension while allowing wrist motion. To date, the amputation levels have allowed for self-contained prostheses either incorporating the battery and controller within the hand or integrating them into the silicone socket above the wrist. The silicone sockets can be designed to house the battery and electronics within pockets created during fabrication process

(Figure 4). Wires are passed across the wrist through channels. Zippers can be included to ease donning/doffing.

PROSTHESIS CONTROL

Control of the fingers has been provided by either single site myoelectric control or FSR using the LTI Varigrip microprocessor. Myoelectric signals have generally been acquired over either the thenar or hypothear musculature. In one case, a C 5-6 quadriplegic was fitted using wrist extensors for control due to the absence of muscle signals within the hand due to his neurological deficit (Figures 10, 11). When physiological thumb motion is available, either with the thumb remaining or when harnessed for control of a post as described above, the hypothenar muscles are used for control. Finger control can be enhanced by varying the on threshold allowing the paired fingers to move in a more natural fashion or to provide specific function. For example, if the on threshold for the ring and pinky fingers is set lower than for the index and middle finger the user can curl the fingers into a flexed position in a more natural way by first producing a small signal, causing the ring and pinky fingers to flex, and then increasing the signal to activate all fingers. Future controllers may be designed to use this feature to provide specific functions.

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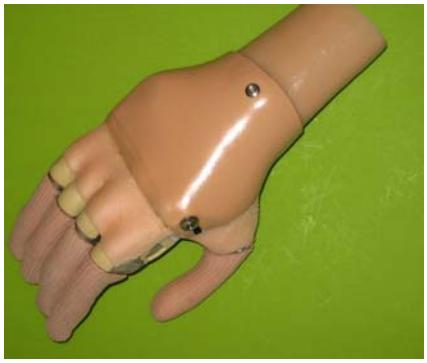

Figure1

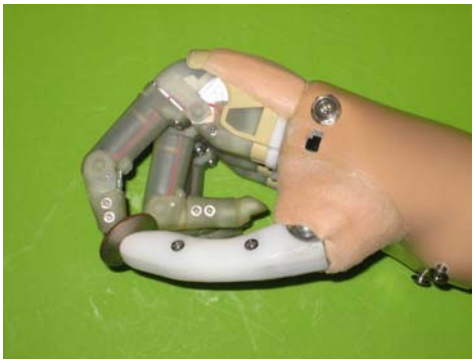

Figure 2

Figure 1 – Finished prosthesis with FSR control using alternate control scheme.
Figure 2 – Friction adjustable thumb demonstrating tip prehension pattern

Figure 3

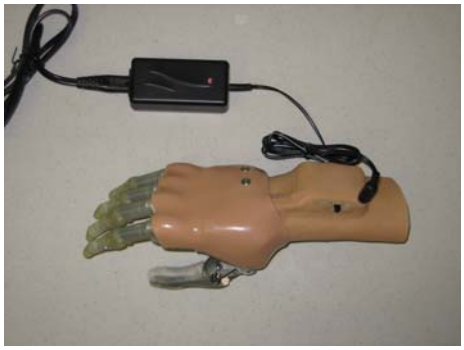

Figure 4

Figure 3 – Powered thumb set up for trial fitting. The thumb mount is attached to the knuckle block in such a way to allow rotation to 45 degrees to provide lateral prehension. The protrusion of the powered thumb was cosmetically unacceptable to the user and was replaced with the Hosmer two position thumb in the final design.
Figure 4 – Finished prosthesis with battery mounted within the silicone socket.

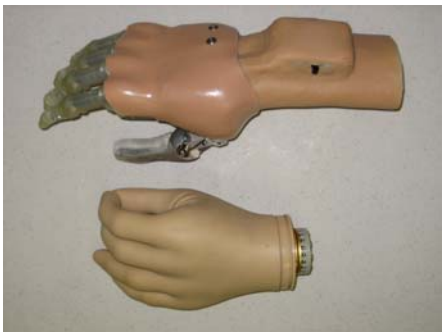

Figure 5

Figure 5 – Finished prosthesis including battery and all components compared to a 8 1/4" hand. Both weigh 505 grams.

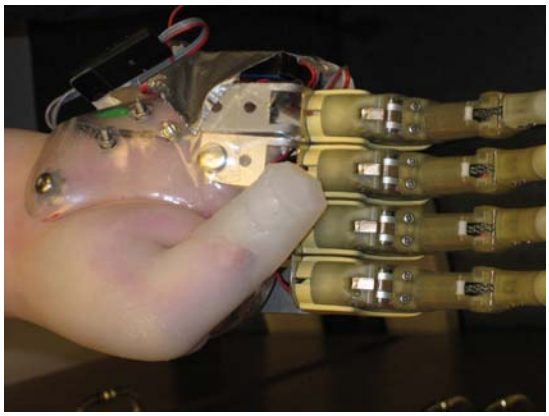

Figure 6

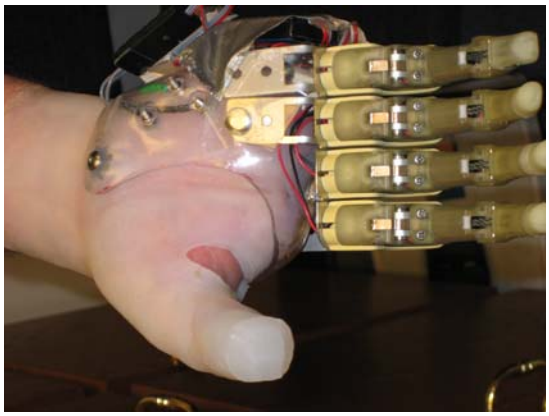

Figure 7

Figure 6 – Prosthesis for trial fitting with partial thumb remaining demonstrating thumb flexion range.
Figure 7 – Same patient as fig 6 showing thumb extension range. The split was intentionally created to reduce resistance to extension, the final design was fabricated with a similar split socket.

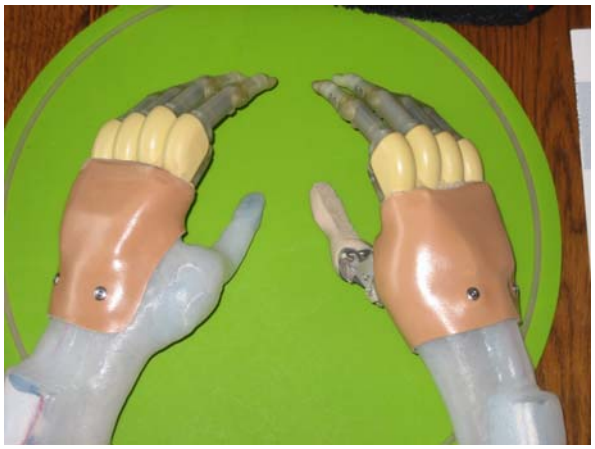

Figure 8

Figure 9

Figure 8 – Prostheses for bilateral partial hand fitting. The right hand utilizes a Hosmer two position thumb. The left prosthesis captures thumb metacarpal motion and allows for force feedback to sensate skin.
Figure 9 – Hosmer two position thumb in wide opening position.

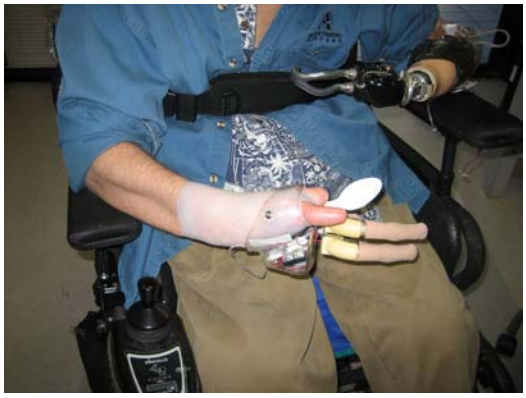

Figure 10

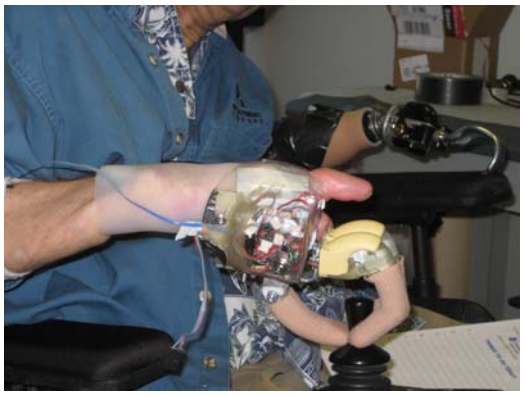

Figure 11

Figure 10 – Fitting for C 5-6 quadriplegic. The residual fingers have no motion. A utensil holder has been incorporated in the custom silicone socket.
Figure 11- Two powered fingers in opposition to a friction thumb provide a three point grasp. Control is achieved by myoelectric control using forearm extensors due to lack of active muscles within the hand.

Cable Driven Multi-Articulating Fingers, Providing Compliant Grasp For The Partial Hand Amputee.

Matthew J. Mikosz

This article will describe a new concept in fitting the partial hand amputee to offer them enhanced grasp and improve their overall functional ability with the prosthesis. The concept is to activate the mechanical fingers through existing wrist motion to provide active grasp that is compliant to any object. A passive thumb was used that allowed frictional positioning of both flexion/extension and rotation of the thumb. The mechanism that pulls the fingers into flexion provides an even distribution of pressure throughout each finger to ensure a secure grasp on the object being held. This concept was designed to offer the metacarpo-phalangeal level to trans-carpal level amputee the ability to achieve active grasp with their prosthesis and enhance their overall functional ability.

Title: **Angulation Osteotomy to Improve Function in Transhumeral Amputee Rehabilitation**

Presenter & Primary Author: Troy Farnsworth, CP, FAAOP
Hanger Prosthetics and Orthotics
Vice President, Upper Extremity Prosthetic Program

Co- Authors: Del Lipe, CPO
John Fergason, CPO
Robert Granville, MD
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Learning Outcomes: At the conclusion of this presentation the attendees will:

1. Understand the clinical basic surgical considerations of a angulation osteotomy
2. Understand the prosthetic considerations of a angulation osteotomy
3. Understand the rehabilitation considerations of a angulation osteotomy

Abstract:

Regardless of prosthetic design, there exist inherent limitations which adversely affect the functionality of transhumeral prostheses. Loss of voluntary humeral rotational control, limitations in prosthetic suspension and decreased range of motion limit the user's acceptance and functional use of a prosthesis.

Various techniques have been discussed in the medical literature to compensate for these shortcomings. These include socket design techniques, harness techniques, and surgical techniques.

Marquette introduced the concept of humeral angulation osteotomy to resolve these issues. By surgically angling the distal humerus the amputee can be fit with a self-suspending prosthesis that enables voluntary rotational control without restrictions to range of motion. Although this technique is discussed in various prosthetic text books, very few cases have been reported. Case studies will be presented showing surgical, rehabilitation, and prosthetic considerations.

Angulation Osteotomy to Improve Function in Transhumeral Amputee Rehabilitation (page 2)

Presenter's experience:

Presenter has extensive clinical expertise as well as lecturing experience including:

AAOP Upper Limb Fellowship Module
AOPA presentations
AAOP presentations
ISPO presentations
MEC presentations
CMSA presentations
PM&R presentations

Biographical Sketch of Presenter:

Troy Farnsworth, CP, FAAOP is an American Board for Certification accredited prosthetist and licensed engineer, specializing in upper extremity prosthetics. As the Vice President of Hanger Prosthetics & Orthotics, Inc. National Upper Extremity Prosthetic Program, Mr. Farnsworth lectures, educates and demonstrates to patients, therapists and physicians nationally and internationally. Clinically he specializes in the rehabilitation of difficult cases of upper limb loss. He currently provides clinical consultation and services for the Center for Intrepid at Brooke Army Medical Center (BAMC) in San Antonio, Texas for amputee soldiers returning from global conflicts.

Title: **Enhanced Functional Outcomes with Elective UE Amputations**

Presenter & Primary Author: Troy Farnsworth, CP, FAAOP
Hanger Prosthetics and Orthotics
Vice President, Upper Extremity Prosthetic Program

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Abstract:

With modern limb salvage procedures, successful reconstruction is not always achieved. This lecture will utilize case presentations to show increased functional benefit in patients who underwent elective amputation surgeries. Neurological damage, severe burn, and brachial plexus injuries are common occurrences that lead to a non-functional problematic reconstructed limb. Typically considered failures by the surgical team, potential prosthetic users need to be made aware of the positive benefits of elective surgeries. Preliminary outcomes measures will be presented including South Hampton Hand Test and Functional Capacity Evaluations.

Presenter's experience:

Presenter has extensive clinical expertise as well as lecturing experience including:

AAOP Upper Limb Fellowship Module
AOPA presentations
AAOP presentations
ISPO presentations
MEC presentations
CMSA presentations
PM&R presentations

(Page 2) --- **Enhanced Functional Outcomes with Elective UE Amputations**

Biographical Sketch of Presenter:

Troy Farnsworth, CP, FAAOP is an American Board for Certification accredited prosthetist and licensed engineer, specializing in upper extremity prosthetics. As the Vice President of Hanger Prosthetics & Orthotics, Inc. National Upper Extremity Prosthetic Program, Mr. Farnsworth lectures, educates and demonstrates to patients, therapists and physicians nationally and internationally. Clinically he specializes in the rehabilitation of difficult cases of upper limb loss. He currently provides clinical consultation and services for the Center for Intrepid at Brooke Army Medical Center (BAMC) in San Antonio, Texas for amputee soldiers returning from global conflicts.

CAPTURING SHOULDER MOTION AS AN INPUT FOR EXTERNALLY-POWERED, SHOULDER DISARTICULATION PROSTHESES

Robert D. Lipschutz, Jonathan Sensinger, Blair Lock, Todd A. Kuiken

Prosthetists have been fitting externally-powered components to individuals with “shoulder disarticulation”, upper extremity amputations for decades. These components have ranged from momentary contact switches that permitted carbon-dioxide to pass through tubes in order to create an articulating motion, to force sensitive resistors (FSRs) that vary the amount of resistance between thin conductive plates in order to permit varied current to flow and provide input to an electrical motor. Activation of these types of inputs requires contact by the user with their remaining residual limb or, in the case of individuals with congenital deficiencies, phocomelic digits. A variety of pull switches have also been used to harness the body motions provided by the user, which activate an electro-mechanical switch used to drive a motor. With the use of coupled rotary potentiometers, the authors have chosen to investigate a unique approach to ipsilateral shoulder motion as a control source for two degrees of freedom.

Individuals with shoulder disarticulation amputations present major challenges for functional prosthetic restoration. Conventionally controlled, cable-driven prostheses for shoulder disarticulation and humeral neck amputations require the use of scapular or bi-scapular protraction. This requires that the individual have good shoulder and torso posture and maintain adequate range of motion within the prosthetic socket(s). The absence of scapulo-thoracic and sterno-clavicular motion in intra-scapulo-thoracic level amputations make individuals with said amputations inappropriate for the device being considered in this study. It is well know that 2 inches of excursion is required to fully open an adult, voluntary-opening, split-hook terminal device while another 2 ½ inches of excursion is required to fully flex a mechanical elbow (provided the elbow flexion attachment is placed in the original, “starting position”). Mechanical analysis of these systems can easily prove that changes to reduce the necessary excursion are possible by moving the location of the actuation lever or elbow flexion attachment; however, these changes are minimal and require a trade-off of increased force requirements. Components such as excursion amplifiers have traditionally been used to enable the users of such body-powered prostheses to have the necessary excursion to control the prosthesis with these aforementioned increases in force.

This being said, unless highly motivated, many individuals with these higher levels of amputation will discard their body-powered prostheses due to the limited tangible benefits of the device. Scapular protraction is being used to activate elbow flexion and/or terminal device operation. This gross body movement does not physiologically translate into the actions being performed by the prosthesis and compounds the drawback of increased force requirements. Although the motions used in this study are not completely physiological; they have closer ties to the functions for which the users intends the prosthesis to perform.

Due to the aforementioned limitations of completely, body-powered prostheses for individuals with shoulder disarticulation amputations, externally-powered or hybrid prostheses are often recommended. The use of externally-powered components decreases the necessary excursion and force requirements by the user and may provide benefit of increased joint range of motion and/or grip strength. For obvious reasons of epidemiology, the focus of externally-powered

components has been in the area of elbows, wrist rotators and terminal devices. There has been little focus on the development of externally powered humeral rotators or shoulder joints.

Humeral rotation, in upper extremity prostheses, has been provided primarily by means of passive pre-positioning. Turntables or other components permit motion via creation of an external moment of internal or external rotation with respect to the elbow axis in the transverse plane. This is usually performed with the contralateral hand or against the body or stationary object (in the case of individuals with bilateral upper extremity amputations). There exists a locking/free-motion, humeral rotator available that permits re-positioning of the elbow in the transverse plane through a sophisticated cabling and harnessing control. Although this is very effective in providing an increased independence to the prosthetic user, it requires significant motivation and training in order to perfect the control motions without expending large amounts of effort.

Mechanical shoulder joints of 0 through 3 degrees of freedom have existed for many years. Monolithic construct provides for an aesthetic appearance at the shoulder without function. Single-axis joints, usually set in the coronal plane for shoulder abduction and adduction provide the user the ability to more easily don clothing, etc. These are often reserved for individuals with humeral neck amputations to provide some motion without creating large asymmetries in hemi-shoulder widths from the midline of the body to the acromial region. The most popular shoulder joints are the bi-axial shoulders that have motions of flexion and abduction. They provide the same benefits as the single-axis joints and also provide for sagittal plane motion. This sagittal plane motion is a necessity for users to perform bimanual activities or reaching tasks either in front of the body or overhead. Although tri-axial shoulder joints exist, they are used infrequently. This is due to the fact that, as mentioned above; many of the elbow units that are used in conjunction with mechanical shoulder joints have transverse plane motion available passively through a turntable or other friction joint. A third plane of motion at the shoulder is; therefore, unnecessary. The tri-axial, or universal joints, are mainly used for passive prostheses when the user wishes to have free-swinging shoulders for aesthetic appearance at rest or during ambulation and are rarely used with mechanical friction settings that would enable stationary re-positioning or pre-positioning for use of distal components. Lastly, there exists a shoulder that has the ability to lock/free swing in one plane. This shoulder joint incorporates passive abduction with the locking and free-swing capabilities in the sagittal plane. Individuals with unilateral amputations use the free-swing feature because it permits comfort between the socket and limb interface during ambulation. Individuals with high, bilateral amputations will use these joints for re-positioning of the shoulder for a variety of activities. This re-positioning is performed by unlocking the shoulder joint, using gravity combined with forward trunk lean to place the shoulder joint in the desired location, re-locking of the shoulder joint and return to an upright, trunk posture. This shoulder joint has the ability to greatly enhance the functional abilities of the user, especially those with bilateral amputations. Drawbacks of this joint include the precise placement of the mechanical switch to lock and unlock the shoulder and the gross body movements that are necessary to re-position the joint, although the former may be remedied with an electronic lock/unlock strategy.

Limitations of mechanical shoulder joints were the impetus to investigate the use of a powered shoulder joint and corresponding control schemes. Three electromechanical control strategies

were tested: a pair of FSRs combined with a rocker switch, three linear transducers and a two-axis joystick. Initially, testing was performed on able-bodied individuals to determine the amount of available shoulder excursion during two antagonistic pairs, elevation/depression and protraction/retraction. Measurements were obtained with the use of an "oversized" shoulder cap mounted on an adjustable stand and a telescoping rod and sheave. The shoulder cap was placed around the subject and the rod was placed on strategic locations for measurement of these motions. Acromial travel for elevation and depression was measured using the telescoping rod placed laterally over the acromial area. Subjects were asked to first elevate and then depress their shoulder while measurements were taken between trials. The same procedures were performed anteriorly and posteriorly at the humeral head and distal scapula respectively. Each subject performed three trials of each motion and the data was recorded. Average amounts of travel were calculated for four subjects. These distances were used to determine how much travel could be applied to the joystick and where the attachment, along the axial distance of the joystick, had to be mounted. Although the amount of travel for the two antagonistic pairs was not symmetrical, this could be accounted for by altering thresholds and sensitivities in the programming software.

Bypass sockets were fabricated for three able-bodied individuals in order to test the three control strategies. The first strategy that was tested was a configuration that had been used clinically for several individuals with shoulder disarticulation amputations. Inside of the bypass socket was a combination of two FSRs mounted anteriorly and posteriorly on the lateral portion of the socket with a rocker switch mounted (upside-down) superiorly. (figure 1) The FSRs were used for control of internal and external rotation of a powered humeral rotator. The rocker switch, was used to control powered shoulder flexion and extension. The second strategy was designed with three linear transducers tethered to a mobile shoulder cap. Initially, the shoulder cap articulated with the socket with a urethane ankle joint. This configuration proved to be too rigid and thus the shoulder cap was loosely tethered by elastic webbing while the cords attached to the transducers themselves, along with the anatomical contour of the plastic, held the cap into place. (figure 2) Lastly, a two-axis joystick was mounted to the medial portion of the bypass socket while the loose cap was modified with an eyelet mounted superiorly to accept the extended rod of the joystick. It was essential that the eyelet permit the rod to slide in order to prevent binding and inadvertent signals.

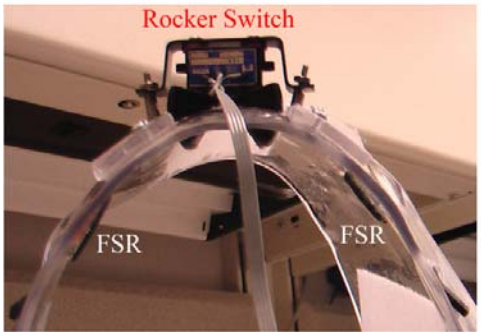

Figure 1

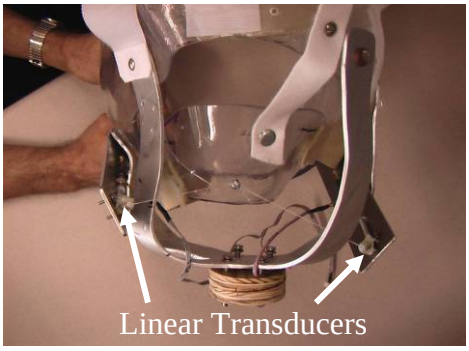

Figure 2

Subjective feedback from the three strategies was obtained. Able-bodied individuals experienced similar problems, with the first strategy, to that worn by the individuals with shoulder disarticulations. Attempts to locate the FSRs and the position of the rocker switch with their lateral shoulder proved difficult. Additionally, the subjects often contacted FSRs

inadvertently while attempting to locate the rocker switch and vice versa. Balancing the three linear transducers proved the most challenging. Antagonistic pairs were set anteriorly and posteriorly for control of the humeral rotator while a single transducer was used for shoulder flexion and extension. A neutral location between end ranges of one transducer had to be found for a "resting state" so that elevation and depression could independently control the two powered shoulder motions. Joystick control proved the most effective as the subjects felt that it enabled them to move more naturally with motions that were somewhat intuitive to the powered motions of the shoulder and humeral rotator.

Following the able-bodied trials, two individuals with shoulder disarticulation amputations were able to utilize the joystick control in experimental fittings. Both of these individuals had undergone targeted muscle reinnervation and thus had independent, myoelectric control of elbow flexion/extension and terminal device open/close. The shoulder controller was added in order to provide inputs for externally powered shoulders and humeral rotators. Two styles of powered shoulder joints were used. One shoulder had powered flexion and extension only, while the other had the addition of powered abduction and adduction. Two, similarly designed, powered humeral rotators were tested as well. The first prosthetic design using the joystick incorporated shoulder elevation and depression for powered shoulder flexion and extension while shoulder protraction and retraction were used for powered humeral internal and external rotation. The second prosthetic design incorporated protraction and retraction for powered shoulder flexion and extension while shoulder elevation and depression controlled powered shoulder abduction and adduction. While the mobile, shoulder cap proved effective for the able-bodied subjects, hollow tubes mounted to form fitted shirts, on one subject, and a thin elastic band on the other (figure 3) proved even more effective in capturing the intended movements of the user. Furthermore, mounting of the joystick laterally created a mechanical excursion advantage which required less shoulder motion in order to drive the powered shoulder or humeral rotator.

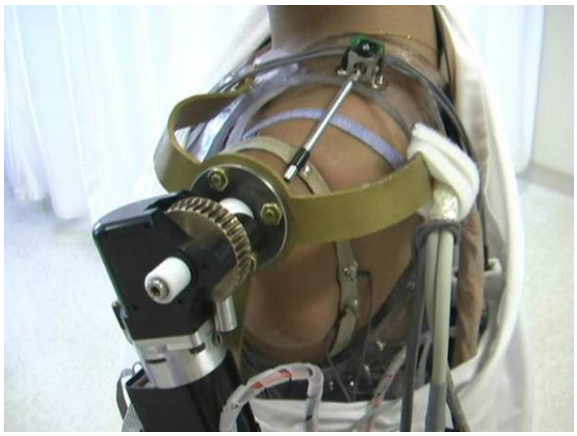

Figure 3

While the subjective feedback from the able-bodied and shoulder disarticulation subjects proved very positive for the joystick design, it is yet to be utilized in clinical fitting. The experimental trials on subjects with shoulder disarticulation amputations were such that the commercially available components (i.e. elbow, wrist and hand) were controlled via EMG control. This is due to the fact that they had undergone targeted muscle reinnervation procedures. There should be little hesitation to preclude the use of this device for individuals with this level of amputation in order to provide control sources for commercially available and experimental components.

BIOMECHANICAL ANALYSIS IN ARM PROSTHETICS – OBJECTIFYING OF FUNCTIONAL ADVANTAGES OFFERED BY WRIST FLEXION

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ABSTRACT

By means of flexion wrists the functionality of prosthetic fittings for upper extremities can be improved significantly. Objects can be manipulated more favourably. Furthermore an optimized gripping position allows an improved body posture. Reaching objects or other dynamic processes are essential in amputee's ADL's. The path of motions influences the whole extremity in a physiological manner.

Biomechanical measuring results demonstrate the advantages in movement by individual hand positioning. Measurements were conducted by means of an optoelectronic camera system for recording motion kinematics (VICON 460, VICONPEAK Oxford, GB). The subjects were fitted with myoelectric arm prostheses and performed predefined motion tests. Position of flexion angle and hand rotation varied systematically. Depending on the posture of the hand the optimization of certain motion patterns could be proved. Repeating action is supported and can be performed more persistently. Some activities confirmed the significance of optimized interaction between wrist rotation and flexion.

INTRODUCTION

Benefits offered by arm prostheses are increased with each additional degree of freedom. A flexion wrist allows optimized gripping and manipulating of objects in different positions required in many daily situations. In a motion analysis lab it shall be visualized how the motion pattern of the whole arm is physiologically influenced.

MATERIALS AND METHODS

Subjects

Six subjects with below arm amputation or malformation were integrated in the tests. As prosthesis the flexion wrist MyoWrist Transcarpal 10V38 or the MyoWrist 2Act 10V40 was used in combination with a myoelectric prosthetic hand. These wrist joints permit locked flexion and extension of 0°, 20° and 40°. It was not allowed for the prosthetic device to cause additional unusual excess length of the system.

Instrumentation

The measurements were conducted by means of an optoelectronic camera system recording motion kinematics (VICON 460, VICONPEAK Oxford, GB). According to the VICON Upper Limb Model passive markers (in total 25) were placed on the upper part of the subjects' body (figure 1) [1]. Based on the configuration of these markers the 3D angles of shoulder, elbow and wrist joint are calculated following the Euler convention. In addition the coordinates of the markers allow identification of important compensation movements of the upper body part using external software (MS EXCEL).

Measurement procedure

At first the single motion patterns were twice performed with the intact side followed by the same motion pattern with the prosthetic side. These movements were repeated six times. If the goals were not reached, only three motion cycles were recorded. In each test, the starting position was standing posture with hanging arm in neutral-0-position. The prosthetic hand was closed. The following motion patterns were performed at moderate speed without physical effort:

1. Hand to the sternum

The subject touches a defined point in the middle of the sternum. Afterwards the arm was moved back into initial position. This movement imitates activities in the middle of the body, e.g. buttoning of shirt or using of zipper.

2. Hand to the contralateral spina iliaca anterior superior

The subject touches the contralateral spina iliaca anterior superior. Afterwards the arm is moved back into initial position (see figure 1). This represents activities at the contralateral side, e.g. threading in of girdle or tucking the shirt into the trousers.

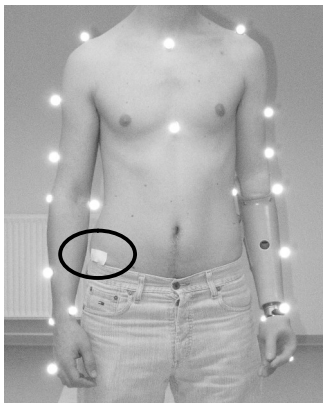

Figure 1.
Patient motion pattern 2

3. Hand to the ipsilateral rear trouser pocket [2]

The subject touches the ipsilateral rear trouser pocket. Afterwards the arm is moved back into initial position. This motion simulates activities at the rear body side, e.g. taking wallet, personal hygiene.

4. Hand to the mouth

The subject is standing in front of a table. The table board is level with the spina iliaca anterior superior. On a defined position a mug half filled with water is placed. The subject takes the mug and moves it to the mouth. After having drunk the mug is placed back to initial position. This movement imitates activities near the mouth, e.g. eating, drinking etc.

5. Hand to the head

For this test the subject is standing in front of a tripod. He moves the prosthesis to a defined place level with the head. Afterwards the arm is moved back into initial position. This motion pattern represents activities away from the body level with the head, e.g. driving a nail into the wall, taking objects from a rack or cupboard.

RESULTS

The results are explained using data from motion pattern 2. Figure 2 shows the typical structure of a shoulder angle. The dark graph illustrates anteversion of the intact arm when the marked position on the hip is reached (see figure 1). With a wrist flexion angle of 40° required abduction of the whole arm is significantly reduced compared to prostheses without wrist flexion. In the following figures the amplitudes of joint movements are demonstrated. Increased wrist flexion leads to reduced compensation movements in shoulder and elbow (see figure 3 – 5). Considering the results it is striking that in each joint and level the angles on the prosthetic side are higher than on the intact side. On the unaffected side, shoulder rotation, elbow flexion and wrist flexion are primarily used to reach the marked position (see figure 4 – 6). In anteversion and abduction an unfavourable lever for holding the arm is produced.

These movements are reduced on the intact side. The range of motion is restricted by the prosthesis causing compensation movements during abduction and anteversion primarily in the shoulder.

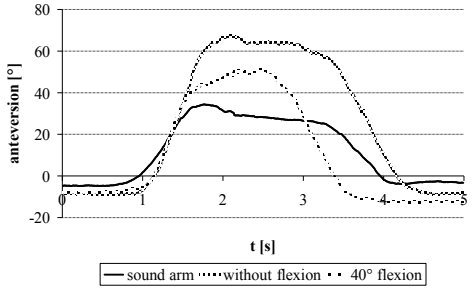

Figure 2. Stucture of the anteversion angle

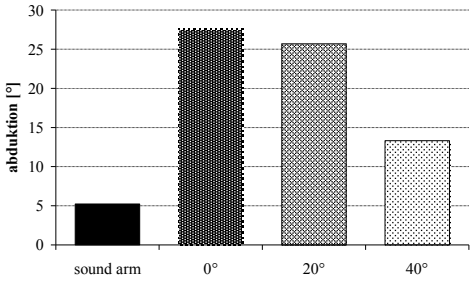

Figure 3. Maximum abduction angle

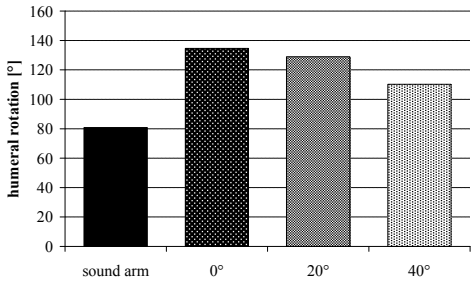

Figure 4. Maximum angle of humeral rotation

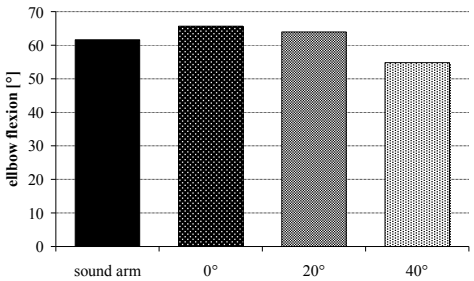

Figure 5. Maximum angle of elbow flexion

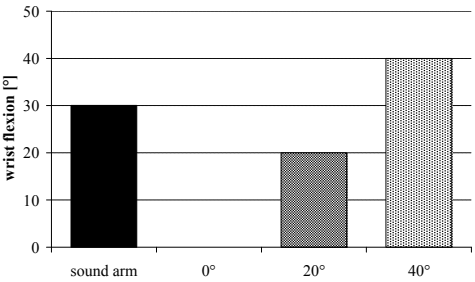

Figure 6. Wrist flexion angle

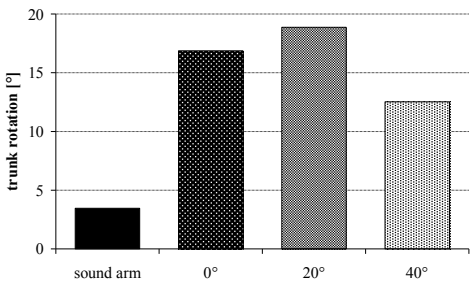

Figure 7. Maximum angle of trunk rotation

To reach the position on the contralateral side additional trunk rotation is required. In this case reduced trunk rotation with increasing flexion angle of the wrist is shown (see figure 7).

CONCLUSION

Different motion patterns performed by different patients were analysed. Before starting the tests hand rotation was adjusted to a position most favourable for the patient. It could be observed that some patients pronated in contrast to other patients that preferred supination although the motion pattern was identical. This can be attributed to different orientations of the sockets. Major compensation movements are generated in the shoulder. However, higher abduction respectively anteversion causes unfavourable arm levers. Further compensation movements are produced by trunk rotation. The results of investigation suggest that flexion and extension of the wrist leads to reduced compensation movements. Combined with rotation compensation movements may be minimized additionally.

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ANALYSIS OF FINGER POSITION DURING TWO AND THREE-FINGERED GRASP: POSSIBLE IMPLICATIONS FOR TERMINAL DEVICE DESIGN

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INTRODUCTION: Analysis of human hand kinesiology has been used to justify the design characteristics of terminal devices. The human hand is a complex mechanical structure consisting of 15 articulations controlled by over 30 muscles for the 5 fingers. It functions as a communication, sensory, grasp and manipulation structure. Separating essential hand kinesiology that is linked with each hand function might allow for the efficient development of terminal devices where there is one primary function such as grasping. It follows that to develop an effective grasping mechanism for a terminal device there should be a clear understanding of hand motions as they relate directly to the grasping function.

The conceptual framework used to characterize grasping functions will most likely influence the development of terminal devices. There is a growing body of literature that is refining the way we think of grasping activities, and may impact the design of terminal devices. For example, researchers have described the simplified coordination of hand motions during grasp as “synergies” or “virtual fingers”, where there are predictable dependencies of one finger on another. Using conceptual approaches such as the “virtual fingers” concept will reduce the complexity of the description of the hand/object/task interaction.[2]

Recent studies indicate that only a small part of the hand’s potential motion is used during grasp. Mason et al. [5] concluded that much of reach-to-grasp uses a base posture with small refinements in finger and thumb positions. In another study Baud-Bovy & Soechting [1] found that when two fingers oppose the thumb, there is a predictable singular task that produced a balanced lever. Kamper et al [3] studied a variety of grasping tasks using 5 everyday objects. They found that the thumb, for example, used less than 5% of the available range. If one were to extrapolate these findings, prosthetic grasping mechanisms may only need a portion of the hand’s total range of motion. Grasping patterns of the normal hand should be studied using a wide variety of objects to compare normal grasp patterns with prosthetic devices. The current study examined finger positions during two- and three-fingered grasp of a wide range of geometrical objects. The results from people with normal hands were compared to two terminal devices.

METHODS: Seven subjects participated in the study (2 male, 5 female). A 6-camera video motion analysis system (Qualisys, Gothenburg, Sweden) was used to collect marker locations in a calibrated volume approximately 1 m^3. Fourteen reflective markers were taped to the interphalangeal joints, metacarpophalangeal joints and the carpals of the subjects (Figure 1). Seated subjects were asked to grasp objects, one at a time, using either two fingers or three fingers. The 9 objects were chosen based on standard geometric shapes, sizes and varied mass distribution properties. The smallest object was a 1.9 cm diameter sphere and the largest object was a 9.5 X 12.5 X 1.7 cm rectangular prism. The objects were painted brown to avoid inferring a particular use. Data were collected at 120

[1] The author has an intellectual property interest in a device discussed in this paper.

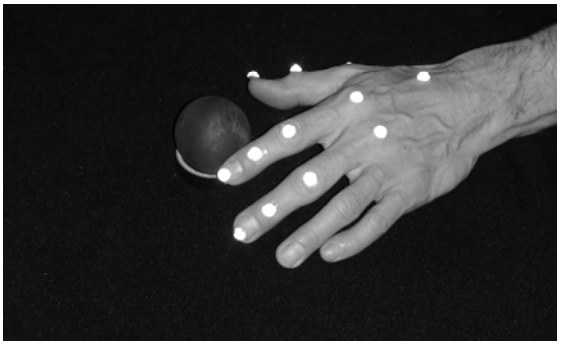

Figure 1. Grasping a medium-sized sphere

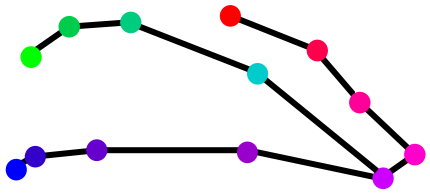

Figure 2. Stick model of grasping a medium-sized sphere

frames/s and the markers' 3-dimensional coordinates were exported for further analysis using mathematical software.

The hand biomechanical model was modified from previous studies of grasp.[4] There were 11 rigid segments generated for the three dimensional motions of each of the finger articulations relative to the wrist (Figure 2). From the three dimensional coordinates of the markers, joint angles and contact angles were calculated.

Pearson product moment correlation coefficients (r) were used to determine the degree of association between distal to proximal joints as well as parallel joints between the fingers. A descriptive comparison was made between motion of the normal hand and two terminal devices (a hook and a modified hook).

RESULTS: Descriptive statistics of joint angles at the time of grasp are listed in Table 1. For two and three-fingered grasp using this specific set of objects, the thumb total range of motion was from 20^{o} to 43^{o} for 80% of grasps. In other words, the range of motion requirement for 80% of grasps in this sample was 23^{o}. The largest variability of any of the finger joints was the index MCP joint (a range of -44^{o} to 49^{o}). There appeared to be a predisposition for the distal segments of the thumb and index finger to be angled rather than being parallel. The average angle between the index and thumb distal segments was about 30^{o}.

Table 1 Joint angle averages, variability and range for all trials and objects Interphalangeal joint (IP), Metacarpophalangeal joint (MCP)

		Mean	Standard deviation	Minimum	Maximum
Thumb	IP	18	10	5	46
	MCP	15	6	3	34
	Total thumb motion	33	14	11	76
Index finger	Distal IP	16	11	1	47
	Proximal IP	24	9	5	39
	MCP	18	28	-44	49
	Total index motion	72	18	38	112
Long finger	Distal IP	19	14	3	49
	Proximal IP	34	9	21	49
	MCP	18	16	-16	53
	Total long finger motion	78	22	35	114

Distal to proximal articulations in separate fingers (thumb, index and long fingers) had relatively low correlations (r= -0.20 to 0.40). Thumb total range of motion had low correlation to index and long finger motion. Higher correlations were found with index finger and long finger parallel joints (r= 0.41 to 0.65) and total

ranges of motion (r= 0.74).

The standard hook is a single hinge system (Figure 3). The hinge allowed for 70^{o} of motion for the distal grasp surface. The alternate terminal device mechanism included the proximal hinge of the standard hook and additional rotation through 4-bar linkages[2] that allow contact surface motion of 90^{o} at each of the prosthetic "fingers" (Figure 4).

Grasp characteristics of the standard hook terminal device depend entirely on the single degree of freedom articulation relative to the size and surface geometry of the object. As a consequence, contact areas are relatively small.

The adapted hook device has a second degree of freedom at each contact surface mechanism that rotates in response to applied forces. The passive mechanisms become tangent to curved or flat object surfaces and can approximate the contact positions of the finger tips while grasping some objects.

DISCUSSION: The available sagittal plane range of motion of index and long finger distal phalanges relative to the metacarpals is approximately 320 degrees and the distal phalanx of the thumb can move 180 degrees relative to the 1st metacarpal, but grasping patterns use only a small proportion of the total range. To develop an effective grasping terminal device, it might be possible to reproduce the necessary range of motion more easily than the potential range of motion.

One goal of a terminal device might be to obtain a similar contact position for the distal phalanx of the thumb, index and long fingers in relation to the objects. For example, the range of motion of a terminal device contact surface should be at least 90 degrees relative to the metacarpals while accommodating to the required aperture.

The single hinge of the hook device can create an aperture for grasp but there is no mechanism to bring the contact surfaces into a more normal finger to surface configuration. The standard hook device is unresponsive to change in forces at object interface. Friction coefficients are low, and in combination with the small surface contact areas and resultant forces on the object that push it away from the terminal device, grasping is less effective.

The adapted hook device has a second level of articulations that allow high-friction material to contact the object. Rotation of the 4-bar linkages allows improved contact geometry. Despite the potential for multiple contact points, the grasp patterns are not similar to the two or three finger grasp patterns of the normal hand. However, adaptations of mechanism may permit a more normal contact distribution.

While it is possible that only the biomimetic

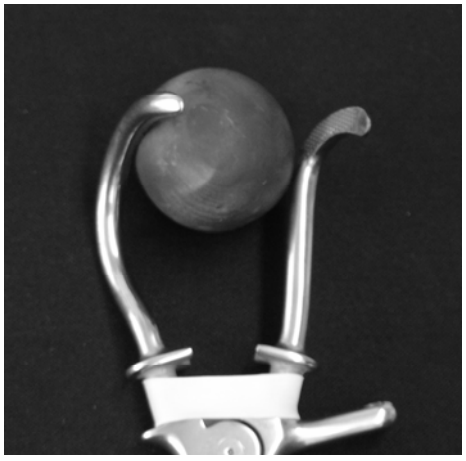

Figure 3 Hook terminal device grasping a sphere

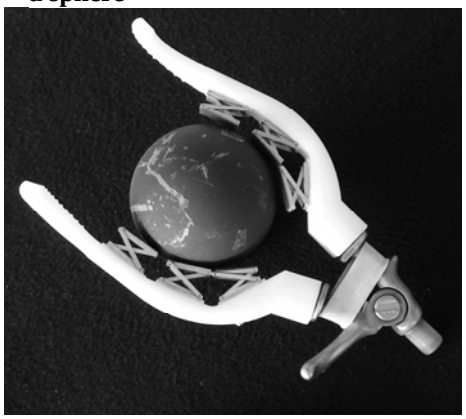

Figure 4 Adapted hook terminal device with 4-bar linkage components

[2]Patent pending

approach can accomplish all possible grips, other approaches with simpler controls or mechanisms have not been explored thoroughly. In creating an artificial hand for grasping, the optimal configuration may differ from the anatomical configuration. Articular designs may lead to contact geometry that is similar to the normal hand.

CONCLUSIONS: Thumb position during two and three finger grasp of standard geometric objects was relatively consistent utilizing less than half of the available range of motion. Most grasps were accomplished within a 20^{o} range for the thumb. The index finger and long finger also used less than 50% of their available range of motion. Design criteria for prosthetic terminal devices for grasp should be derived from normal functional ranges of motion and finger positions used during the grasping process.

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CLINICAL EXPERIENCE AND REHABILITATION OF AMPUTEE MILITARY SERVICE MEMBERS AT THE CENTER FOR THE INTREPID AT BROOKE ARMY MEDICAL CENTER (BAMC): PART 1 – OCCUPATIONAL THERAPY REHABILITATION SERVICES

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Jennifer Menetrez, MD, Center For the Intrepid, Brooke Army Medical Center

This lecture will discuss the clinical experiences and provision of rehabilitation services for amputee military service members at the Center For the Intrepid located at Brooke Army Medical Center (BAMC) Fort Sam Houston, Texas. This is part one of a series illustrating the prosthetic services, rehabilitation and research currently underway at BAMC. Case presentations will be utilized to illustrate the team approach to provide clinically appropriate amputee rehabilitation. This cohesive effort includes surgical considerations, early prosthetic fittings, long-term prosthetic care, rehabilitation services, case management, physiological and social support services. Protocols for advanced prosthetic training will be presented with emphasis on various prosthetic componentry. Advancements in the rehabilitation of the military amputee at the Center For the Intrepid will be discussed in detail to include functional capacity evaluations, drivers training, and firearms training simulator. A case study that involves an individual who sustained a dominant upper extremity transhumeral amputation with total loss of vision will also be discussed. In addition, current and future clinical research studies that involve individuals who have sustained upper extremity loss and who are undergoing rehabilitation at the Center For the Intrepid will be highlighted.

Current war fighting tactics in Iraq and Afghanistan, specifically the use of improvised explosive devices (IEDs), inflict traumatic injury on vast numbers of American and coalition forces [1]. However, technological advancements in medicine, military protective gear, and evacuation transportation have yielded a lower United States casualty percentage in comparison to past United States war conflicts [2]. Therefore, more soldiers are surviving traumatic injuries resulting in limb amputation [4].

Upper extremity amputation often results in significant loss of independence [3]. Soldiers with upper extremity amputations face complex challenges not only in rehabilitation programs but also when performing advanced activities of daily living (ADL) [5]. As of May 2, 2008 there have been a total of 29,911 United States casualties during Operation Iraqi Freedom (OIF) and 1,937 United States casualties during Operation Enduring Freedom (OEF). There have been a total of 770 OIF/OEF service members who have sustained amputations. Of that amount 22% (170 service members) sustained upper extremity amputations [6].

In early 2007, the United States military established the Center For the Intrepid at Brooke Army Medical Center, Fort Sam Houston, Texas. The mission of the Center For the Intrepid is to provide the highest quality of comprehensive outpatient rehabilitation for eligible patients in a

state of the art facility that is unlike any other in the world, facilitating the greatest opportunity for functional improvement and maximal performance in both the military and society while conducting leading edge research, education and training.

The Occupational Therapy rehabilitation service at the Center For the Intrepid is composed of many different rehabilitation specialists who are dedicated to providing the highest level of care to individuals who have sustained traumatic limb loss. Advanced research efforts and the development of new clinical protocols have led to the advancement of function and reintegration of many military service members who have sustained traumatic limb loss. The core of the Occupational Therapy Community Reintegration Program at the Center For the Intrepid is development and enhancement of appropriate psychosocial skills, peer-to-peer interactions, executive functions, motor planning and physical components for individuals with various levels and types of amputations. The overall terminal objective of the Occupational Therapy Community Reintegration Program is to provide a real world environment for individuals with amputations to develop the necessary cognitive and motor skills required for day-to-day social interactions, physical challenges and environmental obstacles.

The Computer Electronic/Accommodations Program (CAP) was initiated to provide assistive technologies and services to individuals with a variety of disabilities and functional deficits that impact the use of information technology and/or job performance. CAP is a Department of Defense (DoD) centrally funded program that provides assistive technologies and reasonable accommodations to individuals with disabilities. The Occupational Therapy process includes the identification of individuals throughout the DoD who could potentially benefit from the procurement and utilization of assistive technology and services. The specific types of assistive technology may include computer voice recognition software and one handed computer devices for individuals with upper extremity amputations.

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Title: **Clinical Experience of Fitting Amputee Soldiers at the Center for the Intrepid at Brooke Army Medical Center (BAMC): Part 2 – Prosthetic Services**

NOTE: Part 1 of the series will be submitted and presented by:
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Abstract:

This lecture will discuss the experiences at the Center for Intrepid located at Brooke Army Medical Center (BAMC) providing rehabilitation of amputee soldiers. This is part two of a series illustrating the prosthetic services, rehabilitation and research currently underway at BAMC.

Case presentations will be utilized to illustrate the team approach to provide clinically appropriate amputee rehabilitation. This cohesive effort includes surgical considerations, early prosthetic fittings, long-term prosthetic care, rehabilitation services, case management, physiological and social support services.

Prosthetic considerations will be presented, including combining the use of technology and basic prosthetic principles. Special attention will focus on unique requirements of traumatic amputations resulting from blast, gunshot, burn injuries, and high demands of the young active users. All advanced systems have been fit at various levels of amputation including fittings with patients who have undergone targeted muscle reinnervation surgery. In addition to being one of the three primary prosthetic amputee care centers, BAMC serves as the primary burn center for the Army worldwide.

(Page 2) --- Clinical Experience of Fitting Amputee Soldiers at the Center for the Intrepid at Brooke Army Medical Center (BAMC): Part 2 – Prosthetic Services

Presenter's experience:

Presenter has extensive clinical expertise as well as lecturing experience including:

AAOP Upper Limb Fellowship Module
AOPA presentations
AAOP presentations
ISPO presentations
MEC presentations
CMSA presentations
PM&R presentations

Biographical Sketch of Presenter:

Troy Farnsworth, CP, FAAOP is an American Board for Certification accredited prosthetist and licensed engineer, specializing in upper extremity prosthetics. As the Vice President of Hanger Prosthetics & Orthotics, Inc. National Upper Extremity Prosthetic Program, Mr. Farnsworth lectures, educates and demonstrates to patients, therapists and physicians nationally and internationally. Clinically he specializes in the rehabilitation of difficult cases of upper limb loss. He currently provides clinical consultation and services for the Center for Intrepid at Brooke Army Medical Center (BAMC) in San Antonio, Texas for amputee soldiers returning from global conflicts.

Experience with Electric Prostheses for the Partial Hand Presentation

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Limb deficiency distal to the wrist represents a common presentation, yet a difficult level to treat with a functional prosthesis. (1) Historically, prosthetic treatment has been limited due to the lack of acceptable electric prosthetic options. Poor results were attributed to challenges including functional limitations of prosthetic technology, patient discomfort, cosmetics and absence of tactile sensation. (2) Until the late 1990s, the lack of acceptable electric prosthetic options as well as concise treatment parameters limited prosthetic treatment. John Michael, M Ed, CPO, FAAOP in the early 90s precisely described the challenge when stating, “The dilemma facing physicians and prosthetists is to determine when our admittedly limited prosthetic armamentarium will add a measure of function to diminish the substantial loss faced by the partial-hand amputee.” (3)

As the specialty of upper limb prosthetics realizes the attention of many different research endeavors, advances in upper limb technology once only found in the research laboratory or in the minds of idealists are beginning to evolve toward commercial availability. These current and future additions to the prosthetic field create a challenge. With the renewed focus on the partial hand level, variables such as residual limb presentation, surgical results, and anatomical stability point to the need for concise treatment parameters.

The Surgical Question – Shifting the Paradigm

As the treatment of the partial hand level progresses, the prosthetist will play a vital role in physician education as to appropriate levels that will maximize the rehabilitation of the individual with upper limb deficiency and in the development of progressive socket designs. Up to this point, hand restoration has been focused on saving as much of the anatomical structure as possible. Truly a worthwhile endeavor, this effort may at times conclude in residual limbs with significant reduced function and problematic pain. While the incidence of pain is more common at the phalangeal level, pain may be realized in more proximal partial hand levels. Ouellette, et al, noted that pain at these other levels may be the “result of an injudicious attempt to save length at all

costs. Although maintenance of length is of concern, such residua seriously jeopardize function of the entire hand." (5) With the ability to apply prosthetic technology effectively at this level comes the absolute necessity for surgical and prosthetic collaboration.

Goals of Prosthetic Management

Our emerging clinical experience point to several specific prosthetic management goals. *Protection* of the residual limb is the first goal. Given the common traumatic presentation of the partial hand level, the residual limb can be significantly compromised in any type of grasping pattern without consideration placed on protection of the residual limb. To address the goal of residual limb protection socket biomechanics must be considered allowing optimal stabilization of the prosthetic socket about the residual limb. One can accomplished this in several methods including custom silicone restoration utilizing a suction fit, non wrist encapsulating or articulated wrist designs which suspend over the residual anatomy, as well as suction type interfaces that extend proximal to and encapsulate the wrist. (4)

The second goal of *bimanual stability* enables the patient to effectively manipulate an object or task using both the sound and affected hand. This goal is directly related to the third goal – *restoration of prehension patterns*. (4) Studies performed both within the prosthetic field and in other clinical areas direct the practitioner's attention to the incidence of contra lateral overuse syndrome. (6) Contra lateral overuse syndrome leads to decrease hand and arm function often associated with pain and discomfort. In some cases contra lateral overuse syndrome can necessitate surgical treatment if conditions that exasperate this condition are not effectively addressed. There are many concurrent modalities that help reduce the prominence of this presentation. Along with medical and therapeutic management, the restoration of prehension patterns and bimanual stability can help promote less reliance on the sound side.

The last prosthetic goal is to provide acceptable cosmesis and durability given the location of the amputation. This goal has proven to be very difficult given multiple and varied joint motion that requires the cosmesis to accept both tensile and compression forces.

Unique Opportunity

Individuals with partial hand amputation represent a patient population that has been previously underserved. The significance of this level can be found in an analysis of the epidemiology of amputation itself. In 2002, Dillingham et al published an article looking at the epidemiology and recent trends in the United States regarding limb amputation and limb deficiency. In his study, he was able to determine that approximately 18,496 individuals yearly are either born without, lose an upper limb

secondary to cancer or experience some type of traumatic insult that results in upper limb amputation. When these numbers were further broken down, it was found that the vast majority was traumatic and, of significant interest, nearly 17,000 of these amputations were distal to the wrist. What this means from a clinical prosthetic standpoint is that professionals in our field are focusing their efforts on the needs of fewer than 2,000 individuals a year who are at a high level of upper limb deficiency, and this minority of amputees comprises the bulk of our typical clinical experience. Meanwhile, there is a huge pool of partial hand level amputees who would no doubt appreciate and benefit greatly from proper prosthetic options if these options were only available to them.

A unique opportunity exists at the partial hand level as our specialty enters a new prosthetic paradigm where evidence based rehabilitation and sound research practices are expected by both the medical community and reimbursement agencies. The opportunity is significant – we have the chance to formulate the research methods and clinical protocols from the ground up instead of retrospectively rationalizing clinical facts that lack the research base today's rehabilitation community requires.

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PRELIMINARY FEEDBACK FROM FIELD TRIAL USERS OF THE MOTION CONTROL MULTI-FLEX WRIST

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ABSTRACT

The feedback provided by patients in the early development of components is essential to the transition of new components into mainstream production and ultimately the success of the new components. This feedback often allows the manufacturers to perform essential design changes to the product to improve durability and function. The Multi-flex (MF) wrist is a compliant prosthetic wrist unit that features both locking positions and spring assisted resistance. This article reports on information gathered from questionnaires on the Motion Control MF wrist during a recent field trial. This article will address the features of the MF wrist, data collected from preliminary field trial questionnaires and the activities of daily living performed by a single user while using the MF wrist.

INTRODUCTION

Field trial testing prior to commercial release of new products is an essential aspect of product development in upper extremity prosthetics. The subject of this article is the Multi-flex (MF) wrist manufactured by Motion Control. This paper will specifically address the features of the MF wrist, field-trial questionnaires collected from four subjects using the MF wrist and a single subject demonstrating his work related job activities.

Research papers written about hand use in performing everyday and occupational activities have shown the importance of wrist motion. Studies incorporating many different tasks noted a requirement of 80 degrees for wrist flexion and extension and 60 degrees for radial and ulnar deviation. When eating, 30 degrees of flexion and 30 degrees of radial and ulnar deviation were needed (1). Wrist motions are essential to everyday tasks and work related activities. Current prosthetic technology only incorporates variable wrist flexion and extension with a series of locking positions and friction based radial and ulnar deviations.

The MF wrist is directly bolted to the specific Motion Control terminal device (i.e., a hand or electric terminal device, “ETD”). The MF wrist affords many different passive as well as locking positions. The basic joint in the MF wrist is a universal joint. The universal joint has a spring assist to center the terminal device (TD) and provide resistance in flexion, extension and radial and ulnar deviation in the unlocked position. The MF wrist locks in one axis at neutral and at the extremes in both directions allowing locking in 28 degrees of flexion and extension. Since the MF wrist only locks in one axis, the user will have to select which axis the hand locks in during the initial ordering and installation of the MF wrist. The unlocked axis allows the same 28 degrees of range in ulnar or radial deviation with spring resistance return to neutral. Initially the MF wrist only locked at the extreme ranges of motion. However, due to field trial user suggestions it now has a neutral locking position. The MF wrist, and seal membrane is depicted in Figure 1. The seal membrane itself was an improvement made to the wrist after information gathered from field trials. The seal membrane keeps the mechanism cleaner, drier and protects the life of the springs.

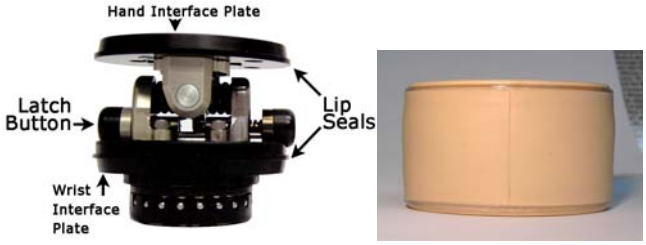

Figure 1: Depicts a close up view of a) MF wrist, latch button and lip seals and b) seal. The latch button engages or disengages the locking mechanism. Proximal and distal lip seals were created to allow the donning of a flexible membrane. This membrane minimizes debris, and moisture entering the MF wrist unit.

Figure 2: Depicts the overall length using an MF wrist compared to the standard hand and ETD

The size and weight comparison for the standard hand and ETD compared to terminal devices utilizing a MF wrist are represented in Table 1. The weight differences noted between a hand and an ETD are due to extra spacers necessary to allow full locking at extreme ranges.

Device	**ETD Std**	**ETD w/ MF**	**Hand Std (sz 7 ¾)**	**Hand w/ MF**
Weight	15.2 oz	17.5 oz	16.9 oz	18.1 oz
Length	6 ¾ in	7 5/8 in	6 in	6 3/8 in

Table 1: Weight differences between the standard hand and ETD, with and without the MF wrist unit.

QUESTIONNAIRES

Upper extremity is a very small patient population; therefore it is often difficult to have large number of field trial participants. Participants for this field trial were selected based on the following criteria: experienced user of an externally powered prosthesis; powered arm is functioning and getting at least 5-10 hours per day use; good communication skills. Although the number of subjects participating is very small, and statistically the data gathered is insignificant, the scope of the information gathered from these field trial users helps guide development and design improvements as well as validates that the design parameters are met.

The questionnaires were administered via telephone from Motion Control. The telephone calls were administered in this fashion so that the questioning could be open-ended and the caller could explore functional areas that a written form may not be able to gather. The ultimate goal of these field trial questionnaires is to gather field trial user feedback to improve function and durability of the product. Any questions about how the wrist is being used in a specific task could be further explored more easily with a phone call discussion.

There were a total of four subjects participating in the field trial tests. The average number of month's usage of MF wrist prior to administration of the questionnaire was 7 months. The amputation level, side of amputation, TD using the MF wrist and date the MF wrist was received by the field trial users are represented in Table 2.

	Subject 1	**Subject 2**	**Subject 3**	**Subject 4**
Amputation level	Long Transradial	Long Transradial	Transhumeral	Transhumeral
Side of amputation	Right	Left	Right	Right
Terminal device using MF	MC hand	MC ETD	MC ETD	MC hand

Date received MF	June 2005	June 2005	January 2006	June 2005

Table 2: Information about the MF wrist field trial users.

The subjects were asked to rate themselves on what type of prosthetic user they are. Three of the four subjects classified themselves as heavy duty users, while one identified themselves as medium user of prosthetic devices. The average hours the subjects reported wearing their prosthesis on a work day is approximately 14 hours per day. On a typical non-work day, the average wear time for their prosthesis was around 12 hours.

The subjects were asked questions relating to the percentage of time they wear the MF TD, percentage of time the MF is unlocked, and percentage of time the MF is locked in flexion or extension, see Table 3. Many of the subjects reported that they prefer to keep the MF in an unlocked position and cited locking the MF wrist only for lifting objects and tool handling. Since the MF wrist utilizes spring assisted return and resistance, many subjects discovered during the field trial usage a desire to have a stronger spring resistance. This suggestion was incorporated in later design improvements. The overall design was improved and simplified to make it easier to replace the springs as well as to install the cover.

	Subject 1	**Subject 2**	**Subject 3**	**Subject 4**
What % are you wearing MF TD?	81-100%	Response not marked	81-100%	81-100%
What % is the MF unlocked?	81-100%	81-100%	81-100%	81-100%
What % is the wrist locked in flexion or extension?	Prefers to keep unlocked	0-20%	0-20%	0-20%

Table 3

The subjects were asked to rate the comfort of the prosthesis compared to previous prostheses. This question was formulated with the idea that creating compliancy with in the wrist could minimize the forces transmitted to the socket making the socket more comfortable. Two of the four rated the comfort as “much better”, one rated as “better” and the last one rated as “about the same”.

The subjects were asked to rate picking up large objects. Three of the four rated this task as worse utilizing the MF wrist. Since the wrist only locks in one axis and the other remaining axis utilizes spring assisted resistance. Holding large objects causes the MF wrist to ulnar deviate.

The subjects were asked to rate ease in performance of tasks. All of the subjects reported “much better” in ease of performance of tasks with the MF wrist. The subjects were asked to rate the naturalness of using the prosthesis. All of the subjects responded with “much better” in the naturalness of using the prosthesis. Each subject rated the overall usefulness of the MF wrist as “much better” as well. All of the subjects reported that the MF wrist allows them to use the prosthesis for more activities. All subjects reported that the MF wrist allows for more grasp than stabilizing. Subjects indicated that tasks where the MF wrist seemed more useful were: handling a knife and fork, riding motorcycles, handling farm sprayer joystick controls more comfortably over longer periods, holding materials down, handling lighter objects such as a can of pop, opening doors, handling kids, driving one handed, holding tools in previously difficult to manage positions, rowing a boat, nailing and handling a fishing reel. Subjects used the MF wrist in the flexed position to pick up heavy objects, ride 4-wheelers, work on small parts in the workshop, cut vegetables in the kitchen, put on a scarf and hold stops in place. Subjects reported advantages of the MF wrist to be more natural, more comfortable, easy to orient for stable grasps and requires less adjustment of position. Disadvantages were reported as accidental locking of

the wrist, unstable when lifting heavy objects, takes time to lock, there is no lock in neutral (corrected through field trial feedback) and cuts gloves.

CASE STUDY

RD is a 32 year old male that sustained a work related injury to the left arm in 2003. The injuries necessitated a right, short transhumeral amputation approximately 3 1/2 inches distal to the axilla level (see Figure 3). Over the course of rehabilitation the subjects residual limb has necessitated several revision surgeries to try and ameliorate pain issues, relieve adherent scarring distally and provide better soft tissue coverage on the distal end of the residual limb.

RD continues to remain active on his family's large crop and cattle farms. He drives large fertilizer (Figure 4) and sprayer vehicles (Figure 5) as well as four wheelers, repairs farm machines in the workshop, and uses drill presses and hand tools (Figure 6). He reports that the compliancy of the MF wrist has improved the comfort of his prosthesis over the course of a day.

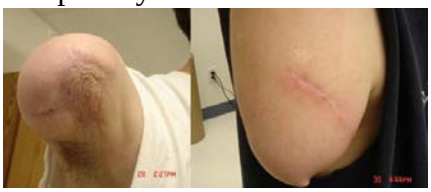

Figure 3: Patient's short transhumeral residual limb characteristics.

Figure 4: Fertilizer vehicle used on the farm. Many of the hand controls found in farming vehicles are located on the right side of the vehicle.

Figure 5: Sprayer vehicle utilized with an average sitting time of about 5 hours. The compliancy of the wrist as reported by the subject has improved his overall comfort while perform work related tasks over the course of the day.

Figure 6: Workshop activities performed to aide in maintenance of farm equipment.

SUMMARY

To fully quantify and understand the user aspects of this wrist a more comprehensive study would need to be performed. However user feedback in the early development of new products is essential in evaluating design goals as well as making design changes that ultimately improve the product prior to commercial release. From the information gathered from the field trial questionnaires we have learned that the users typically preferred to utilize the MF wrist in an unlocked position for 81-100% of the time. They also reported that the ease of performing tasks was improved. Overall comfort of the prosthesis while performing activities was also reported, which may have resulted from the use of a compliant wrist. In summary information gathered from the field trial users aided to improve the design of the MF wrist making it more functional and durable.

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EFFECTIVENESS OF UNDERSHIRT FABRIC ON HARNESS COMFORT IN UPPER EXTREMITY PROSTHETIC USERS: A PILOT STUDY

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ABSTRACT

This pilot study is to investigate alternatives to existing harness/skin interface for upper extremity amputees who use conventional or myoelectric prostheses with a harness. The study design is a single-system prospective study utilizing a population of convenience. A clinic population of upper limb amputees (trans-humeral, trans-radial and trans-carpal) from Tampa Bay Prosthetics and one associated with Northwestern University were included in the study. Participants were provided with two Jockey® Next-to-Nothing™ undershirts to wear as an interface between their harness and skin. They were asked to evaluate the new interface in comparison to their existing wearing pattern. Data for the study was collected using a self-reporting questionnaire that would be filled out after a predetermined wearing schedule. All of the participants were full-time upper extremity prosthetic users. Overall, results from the questionnaire indicates that the new textiles can positively impact the use of a harness in upper extremity prosthetic wearers

INTRODUCTION

The goal of the prosthetist for the upper extremity amputee is to enhance function, appearance, and to make activities of daily living (ADL) possible. Function is not a mechanism that can be measured exactly and is often in a state of flux and specific to the individual.

Both external- and body-powered upper extremity prosthetic users may or will require some type of harness. A harness for an upper extremity body-powered prosthesis has four main objectives: force transmission, suspension, comfort, and cosmesis [1]. Those fitted with body-powered prosthesis will require the use of a harness to aid in the suspension and control of the prosthesis in order to facilitate function of the limb. Often, those wearing a battery-powered or hybrid prosthesis will use a harness for suspension or control of one or more components of the prosthesis, such as the elbow or terminal device.

A variety of harness systems have been developed for specific amputation levels, activity levels and body makeup [2]. Generally, the more proximal the amputation, the more modifications and additions there are to the harness. Operating the prosthesis requires forces to be generated via the harness; this can initiate shear forces between the harness and the skin.

Comfort in the harness is necessary to facilitate maximum functional outcome and long-wearing time of the prosthesis [3,4], greatly impacting the ultimate usage of the prosthesis. Harness comfort was reported as one of the common prosthetic problems associated with prosthetic wear [5-10]. According to Datta et al. [11], as many as 33.75% of patients with proximal upper limb deficiency reject their prosthesis and many of those who continue to wear them do not find them useful in ADL. Abandoning the prosthesis is attributed to pain, weight, no functional benefit and poor cosmesis [11]. Schultz et al. surveyed opinions of prosthetists and other experts in the field to identify those factors that affect successful outcome for upper-extremity prosthetic users. The majority of the respondents considered comfort (socket interface and weight of prosthesis) as the most important factor for unilateral amputees [9]. Additional concerns included function (the most important for bilateral amputees), agility, power and, last, cosmesis [9]. We would like to further explore the relationship between harness comfort and acceptance and increased wearing time of the prostheses.

Traditionally, a cotton undershirt is worn as an interface between the patient's skin and harness. The purpose is to provide comfort, an element of hygiene and to alleviate chafing during ADL. New "tactical" fabrics designed for the sporting industry are available that claim to possess moisture-wicking and friction-reducing capabilities and are breathable. These characteristics may improve harness comfort resulting in better functional outcome.

The intention for measuring functional outcome is to detect clinically important changes that have occurred in response to an intervention [12]. Specific functional outcome measures for adults who use upper extremity prosthetic devices are lacking in the literature [12]. Core measures are aimed at providing a comprehensive measure of different aspects of hand function and prosthetic use [12, 13, 14].

This paper presents an evidence-based single-system prospective study which draws from a population of convenience to evaluate harness comfort using a non-traditional interface. The study utilizes a trial wearing period of tactical fabrics followed by a self-assessment questionnaire used to determine if improved functional outcome can be achieved. Outcomes are measured with scores from quality of life indicators to demonstrate whether or not there is a benefit to wearing these new fabrics by evaluating comfort, moisture-wicking capabilities, friction reduction, and thermoregulation of the body. We devised to measure comfort levels for those patients that are both experienced and new prosthetic users.

Self reporting measures have the potential to be reliable, valid and responsive to change because they provide measures obtained from real world situations that may not be directly observable by a clinician [12].

METHODS

Review of medical records from Tampa Bay Prosthetics identified twenty patients who utilize a harness while wearing their upper-extremity prosthesis. In addition, requests were sent to Northwestern University to have any upper-extremity patients who use a harness and were interested in the study to contact our clinic in order to be included. Introductory letters were sent to all those identified as possible participants inviting them to join in the study. From the twenty patients contacted, nine from our clinic agreed to participate and one from Northwestern University wanted to be included. Information for this study was obtained from medical records, reports by patients, and from a questionnaire that was filled out after a two week wearing schedule of the undershirts provided to the participants. The questionnaire was developed by the author and not formally validated; however, the content was reviewed by colleagues and deemed appropriate for the study question. Additionally, an "Informed Consent" letter was sent to each participant that was to be signed and returned to the clinic.

We had contacted three manufacturers of performance-wear clothing that claim to produce materials that are superior to cotton for their moisture-wicking and friction-reducing capabilities to request donations of clothing for the study. Of those companies, Jockey® responded and provided us with 30 Next-to-Nothing™ undershirts of varying sizes. Two of these undershirts were provided to each participant along with a consent form and a self-administered questionnaire used to obtain data. The questionnaires were composed of 12 questions designed to collected data regarding the experience of using the study fabric as opposed to a cotton undershirt and how it affected comfort. Specific questions related to the moisture-wicking, breathability, and friction-reducing capabilities of the materials. The participants were asked about the type of interface that was normally worn between their harness and skin. Questions pertaining to amputation level and the type of prosthesis the patient wore (myoelectric, hybrid or conventional) were also included. Finally, the participants were asked if they would continue to

wear the study undershirts or go back to their normal routine. If the Jockey® undershirt was rejected at any time during the study, the questionnaire asked for the reason.

RESULTS

So far, Jockey® undershirts and questionnaires were sent to ten participants, of those, four (40%) returned the consent forms and followed-up with the questionnaires. All of the questions were completed in each of the returned questionnaires; these are summarized below.

Demographic details,

Half of the participants in the study had amputations at the transhumeral level and half had amputations at the transradial level.

Prosthetic Usage

Of the patients who returned the questionnaire, two used both conventional and battery-operated prostheses, one used a hybrid and one used conventional only. All of the respondents who participated are full-time prosthetic wearers.

Interface

Of the four respondents, three usually wore a 100% cotton undershirt as an interface between their harness and skin. Of these three, one would cut off the shirt above the waist for comfort. One patient was already wearing a similar undershirt. All respondents wore the Jockey® performance garments for the entire study period.

Comfort

The respondents were asked to rate the overall comfort of wearing the Jockey® undergarment compared to their normal routine. Three provided a response of “much more comfortable” and one responded “no difference”.

Breathability

All of the respondents found the Jockey® undergarment material felt more breathable than what they normally wore under their harness. Three found the test material “much more” breathable, while one found it “more breathable”. One respondent found the breathability to be the greatest improvement of the Jockey® garment over a cotton undershirt.

Moisture-wicking

The moisture-wicking capabilities of the Jockey® undergarment were also rated higher than cotton garments by all respondents. One patient did comment on the questionnaire that this ability caused his outer shirt to become moist, so, this may have actually been a point against the new material’s moisture-wicking capabilities as compared to his cotton undershirt.

Friction-reducing

Patients were asked if there was any reduction in chaffing from the harness while wearing the Jockey® undergarment. Three responded that the friction-reducing capabilities were much greater with the Jockey® garment than with a cotton undershirt, while one responded that there was no difference between the two. Three of the respondents reported that friction-reducing capabilities were the greatest improvement provided while using the Jockey® undergarment.

Continued use

All of the patients reported that they would continue to use the Jockey® undergarment, or one of a similar material, in place of their cotton undershirt.

DISCUSSION

Only 40% of the subjects identified for the study responded with completed questionnaires, for a total of 4 respondents. Half of these have amputations at the transhumeral level and half have amputations at the transradial level. Two of the subjects use both an externally-powered and body-powered prosthesis, one used a hybrid and one uses a body-powered. All are full-time

users. All of the participants found the material in the Jockey® undergarment to provide some benefit over a 100% cotton undershirt; all stated that would continue to wear the newer interface.

Active-wear or "tactical" clothing can positively impact performance in sports and hobbies. Through this study, we have shown that this type of clothing can affect the level of harness comfort for upper extremity prosthetic wearers. It is our hope that this clothing will influence wearing time for the upper extremity prosthetic community when worn as an interface between the harness and skin.

The weakness of this study is predominantly due to the small subject base and the lack of information regarding the respondents who were provided with the undershirts and did not respond; their input may influence the results significantly. It may be that these patients did not return the questionnaires because they discontinued wearing the undershirts, although there was a section on the questionnaire that addressed this issue. This study did provide insight into an inexpensive and readily available avenue for providing increased comfort for some upper-extremity prosthetic users.

Another population to study would be those with upper-limb loss who have rejected their prosthesis, or wear their prosthesis only periodically. We could determine if increased harness comfort would have any impact on their decision to wear their prosthesis.

One method for collecting data could be a web-based, direct-access questionnaire as described by Wright [12] that can facilitate patient completion. The use of a larger network of patients is necessary to incorporate a larger population in order to better determine the effectiveness on new interfaces on comfort aspects associated with the harness.

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PHANTOM LIMB DEVELOPMENT IN CONGENITALLY UPPER LIMB-DEFICIENT INDIVIDUALS

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ABSTRACT

We collected myoelectric data from five congenitally below-elbow limb-deficient volunteers for evaluation of a myoelectric prosthesis control system using contralateral stimulation. The volunteers reported no phantom limb experiences before the data collection, but began to feel phantom-like sensations of their missing hands during the data collection. Some described limits on their new perceptions remarkably similar to the phantom motion limits described by traumatic amputees (i.e., difficulty in fully opening and closing the phantom's fingers). Significant changes occurred in the volunteers' myoelectric data signatures after they began to feel the phantoms.

INTRODUCTION

We collected a large set of myoelectric signals from 10 upper-limb, below-elbow amputee volunteers while they each generated approximately 1000 candidate prosthesis control commands covering 6 motions. Five volunteers were traumatic amputees and five were congenitally limb deficient. Our goal was a large dataset to evaluate myoelectric control concepts, not studying phantom limbs, but we encouraged those volunteers who had a phantom sensation to use it during the data collection [1].

All five traumatic limb-deficient volunteers reported feeling a phantom limb with varying degrees of mobility prior to the data collection sessions. Three of the five traumatic amputee volunteers reported difficulty in completely opening and closing their phantom fingers. One described some difficulty in feeling their wrist. The traumatic amputee volunteers generated the command-specific contraction in their residual limb easily by moving their phantom limb simultaneously in a mirror image of their sound hand's motion. In spite of these apparent limitations, merely attempting to move the phantom limb in unison with the sound limb was sufficient for the traumatic amputees to generate consistent myoelectric signatures from which motions could be distinguished and classified [2].

Table 1: Five volunteers with **congenital** upper limb below elbow limb deficiencies participated in this study.

Subject ID	Sex	Age	Residual Side	Residual Length	Comments
6	Female	32	Right	6.0 cm	Regular body-powered hook user since early childhood. Occasional myoelectric hand use.
7	Male	12	Left	18.4 cm	No prosthesis use since age 6. Cannot fully straighten elbow of residual limb. Some carpal bones present. Residual forearm radius and ulna fused in pronated position. Sound and residual side humerus bones not equal in length.
8	Female	12	Right	7.0 cm	Moderate myoelectric user, hand only. Mother's ultrasound at 14-16 weeks of pregnancy showed two hands developing normally.
9	Male	13	Right	9.6 cm	Moderate myoelectric user, hand only.
10	Male	12	Left	10.4 cm	Occasional myoelectric user, hand only.

Four of the five congenitally limb-deficient volunteers (Table 1) had no phantom sensations at all prior to the data collection sessions (Table 2). The fifth volunteer occasionally felt an itch beyond the end of his residual limb, but did not feel anything like a hand. This matched the conventional wisdom that congenitally limb-deficient people never develop a map of the missing limb in their brains. We expected the congenitally limb-deficient volunteers to develop some conscious mnemonics to generate consistent motion-specific myoelectric signatures, if they could do so at all. We were not optimistic that they could develop distinct signatures for six motions. During the data collection sessions for the congenitally limb-deficient volunteers, however, these volunteers began to feel phantom limb sensations remarkably similar to those reported by our traumatic amputee volunteers.

METHODS

We captured a minimum of four channels of myoelectric data from each limb (both sound and residual limbs) from each volunteer [1]. Electrode sites were based on a quick qualitative assessment of signal strength and variation as the subject attempted some trial motions. During the data collection, the subject sat in front of a PC display monitor on which a drawing of the desired motion (open grasp, close grasp, flex wrist, extend wrist, pronate forearm, or supinate forearm) appeared in random order. The subjects imitated the pictured motion with their sound hand and simultaneously generated the command-specific contraction they had chosen for that motion in their residual limb. We use the term “contra-lateral stimulation” to describe this exercise where the user moves the intact hand and wrist while simultaneously attempting to do exactly the same action with the amputated limb. We recorded at least 150 trials of each motion type per subject. The subject had a brief rest between motion cues, which were given when the subject said they were ready for the next cue. Longer rest breaks and one meal break were given as the subject requested. Total testing time for each subject was only 3 hours.

Prior to our data collection sessions with the congenitally limb-deficient volunteers, we asked our subjects to develop a prosthesis command set (consisting of motion-specific contractions) that they could remember and associate with the proposed prosthesis movements when they saw the motion prompts. All chose to imagine moving a hand on their limb-deficient side through the desired prosthesis motions. We did not discuss the subject's choice of mental imagery until the completion of their data collection session. We discussed perception of a hand on the limb deficient side during the data collection session with only one subject (8), when that subject initiated the discussion.

We recorded the myoelectric signals at 2400 samples per second per channel, using a Fetrode electrode artifact reduction system (UFI, Morro Bay, California) consisting of disposable recessed silver-silver chloride electrodes, an amplifier and a signal conditioner. The Fetrode system has very high input impedance and virtually eliminates induced motion artifact. The recessed, wet electrodes have low motion artifacts relative to those commonly used in prostheses.

RESULTS

After our subjects told us they began “feeling” a limb partway through the data sessions, we looked for the signature of the phantom limb in the myoelectric data and prosthesis controller simulation performance. We used the subject’s comments about when they became aware of the phantom limb plus motion start settling and signal-to-noise ratios to estimate the point at which the “post-phantom” data set began. Three subjects (8, 9, and 10) could name a specific time that

they first felt the phantom—just after an hour's lunch break. The other two (6 and 7) did not pinpoint a specific time, but mentioned an increasing awareness of a sense of having two hands. Subject 6's lunch break came relatively late in the data session. The break and its timing seem to be important in the phantom development.

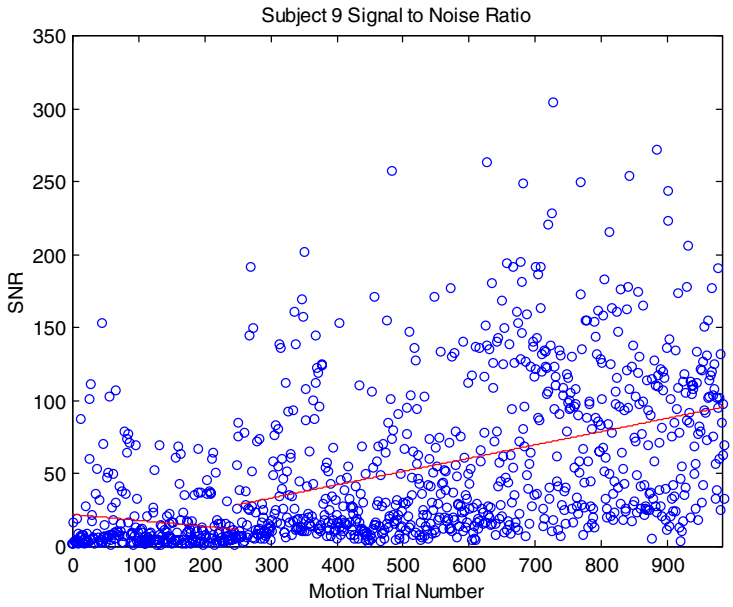
Table 2: Congenitally-limb deficient subject phantom perceptions before and after the data collection session.

Sub-ject	Phantom perception		
	Prior	During	After
6	None	Began by visualizing that she had a hand on her (very short) residual limb and trying to move it in conjunction with her sound hand; could not pinpoint a sudden change or specific onset of phantom sensation. Lunch break occurred just before trial 500.	"After a while, I really got into the feeling that I had two hands." No follow-up was possible.
7	None	Began by visualizing that he had a hand on his (very long, with some carpal bones) residual limb and trying to move it in conjunction with his sound hand; could not pinpoint a sudden change or specific onset of phantom sensation. Lunch break occurred just after trial 400.	Reported a gradually increasing awareness of his residual "forearm muscles pulling at his wrist in a way that made sense." Months later, he could reawaken the sensation by briefly attempting contra-lateral motion.
8	None	Began by visualizing that she had a hand on her residual limb and trying to move it in conjunction with her sound hand; began spontaneously talking about the phantom sensation just after her lunch break. Lunch break occurred just before trial 300.	Reported a distinct change in sensation in her residual limb between the before-lunch trials and the after-lunch trials. Reported difficulty in completely closing and opening the fingers of the phantom hand. No follow-up was possible.
9	None	Began by visualizing that he had a hand on his residual limb and trying to move it in conjunction with his sound hand; reported a distinct change in what he felt on his residual side about 10 trials after his lunch break. Lunch break occurred just after trial 250.	Reported that the sensation came to him as he tried to visualize "a hand opening and closing and moving" just after the lunch break. While the sensation felt like much more than simply the visualization he started with, "it didn't really feel like part of me." Even though he did not feel particularly connected to this hand sensation, he noticed that he couldn't quite straighten the fingers on it. Months later, he could reawaken the sensation by briefly attempting contra-lateral motion.
10	An occasional "itch" located beyond the end of residual limb	Began by visualizing that he had a hand on his residual limb and trying to move it in conjunction with his sound hand; reported feeling like he had a wrist and hand on his residual side just after his lunch break and relying on it to generate contractions instead of the visualization. Lunch break occurred just after trial 399.	He noticed no limitations on the motion of the hand sensation, although its response seemed a little slower than that of his intact hand. Months later, he could reawaken the sensation by briefly attempting contra-lateral motion.

Data analysis revealed four potentially significant quantitative indicators of a "phantom" developing. First, we saw increases in signal-to-noise ratio (SNR) (a measure of strength of contraction, defined as the sum of myoelectric energy on all channels on the residual limb for 0.15 seconds after motion start divided by the sum of myoelectric energy on all channels on the

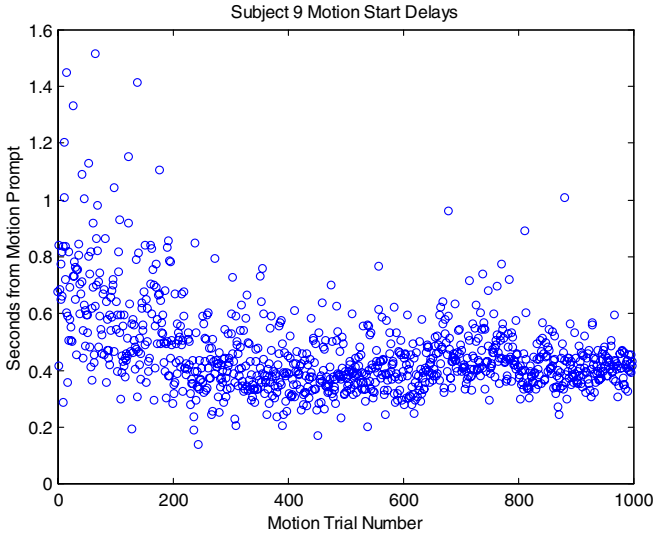
residual limb for 0.15 seconds prior to motion start). On average, SNR improved 250% from the pre-phantom to the post-phantom data subsets. Two subjects ended the data session with steadily increasing SNRs. Figure 1 shows SNR versus motion trial for subject 9, who became aware of a phantom at motion trial 260. SNR slope versus time averaged negative for the five traumatic amputees, as muscle fatigue resulted in weaker contractions over time. SNR slope for our subjects with congenital limb deficiencies averaged positive in spite of fatigue.

Figure 1



Second, the delay between the motion prompt and myoelectric response in the residual limb decreased after the phantom sensation began. Figure 2 shows the motion start delays for Subject 9. The post-phantom consistency is striking compared to the earlier scatter. Third, motion errors (where the subject told us that they made the wrong response to the motion cue) decreased after the phantom perception. Table 3 shows the error rates for the three subjects who had a distinct onset of the phantom sensation. Pre-phantom error rates for these three subjects averaged 4.4% versus 1.7% post-phantom [3].

Figure 2



Fourth, we saw increases in motion identification accuracy of a simulated prosthesis controller [4] after the “phantom” appeared. For two of the subjects (8 and 9) who had distinct onsets of the perception of the missing hand, we compared a genetic-programming prosthesis controller based on (1) the entire data set and (2) the data collected after the subject perceived the "phantom." Evolutionary computing techniques such as genetic programming generally produce better results with larger training data sets, so we expected the accuracy for subjects 8 and 9 to be lower for the more restricted training data sets, all else being equal. Instead, the prosthesis controller motion identification

accuracy increased 3.1% for subject 8 and 4.2% for subject 9. Thus, the contractions recorded in the residual limb were more motion-specific after the subject began feeling the missing hand.

Table 3: A comparison of motion error rates before and after the subject began to feel the phantom sensation

Subject	Total Error Rate	Pre-Phantom Error Rate	Post-Phantom Error Rate
8	2.9%	5.0%	2.0%
9	1.8%	3.6%	1.2%
10	2.9%	4.5%	1.9%
Average	**2.5%**	**4.4%**	**1.7%**

DISCUSSION AND CONCLUSIONS

Is this sensation our congenital volunteers reported truly a phantom limb? Or just very active imagination and increasing proficiency in mental visualization with practice? The myoelectric data suggest that muscles were coordinating in more distinct motion-specific patterns after the phantom sensation began.

The congenitally limb-deficient subjects reported that they felt that both the visible hand and the phantom were moving identically, although perhaps a little slower. The only exception to this was some difficulty in completely closing or straightening the phantom fingers reported by two subjects. The limits to mobility of the hand perceptions—the difficulty in completely closing or opening the fingers—reported by these two volunteers is a particularly compelling qualitative argument that these sensations were a phantom, because three of our five traumatic amputees reported feeling this same limit in their phantoms' mobility.

These results were an unexpected product of a project focused on developing a prosthesis controller. They are preliminary and need validation in a study focused specifically on the development of phantom limbs among congenitally limb-deficient persons.

Acknowledgements

Thanks to T. Walley Williams III, Volunteer Test Subjects, Robert Abramcyzck, Mara Novy, Diane Atkins, Dr. William Donovan, Jeffrey Graham, J. Jaime Fernandez, NASA/JSC, and The Institute for Rehabilitation and Research (TIRR).

Funding

(1) Cooperative Research Agreement between the National Aeronautics and Space Administration's Johnson Space Center (NASA/JSC) and The Institute for Rehabilitation and Research (TIRR) at the Texas Medical Center; (2) , and US Department of Education Small Business Innovative Research (SBIR) ED-98-P0-4645; and (3) National Institutes of Health (NIH) SBIR 1 R43 HD36928-01..

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APPLICATION OF EXTERNAL POWER IN BRACHIAL PLEXUS INJURY MANAGEMENT: A CASE STUDY

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This paper will describe the process of applying external power technology in the traditionally complex clinical area of brachial plexus injury (BPI) management. The acute and long term management of people who have acquired BPI's often presents the clinical team with many complex management challenges. Simpson [1] concisely summarised the core functional prerequisites that are needed for upper limb function as:

- Proximal stability – shoulder integrity
- Placement in space – ability to position the limb
- Functional grasp – the effectiveness of prehension.

When BPI's are so frequently accompanied by closed head injuries, a further prerequisite is simplicity in design and operation. We intend to illustrate the importance of considering these principles through the presentation of a single case study of a client who we have been working with for the past 7 years.

CLIENT HISTORY

Our client is a 43 year old man who had a motorbike accident in 2001. He sustained a right brachial plexus injury (C6-T1) and surgical repair of the plexus was not deemed possible. As a result he presented with an insensate and flail arm with associated shoulder subluxation. He has also had ongoing problems with severe root avulsion pain. In addition, the client sustained a closed head injury (CHI) at the time of the accident, impacting on his memory, concentration and ability to manage stress. At the time of his injury he was married with a 12 year old daughter, and was working as a maintenance engineer that included using lathes and hand tools, and his leisure activities were hunting, fishing and gardening. He was referred to our service approximately one year post-injury.

ORTHOTIC HISTORY

A standard arm sling was fitted early and continues to be used periodically to position and protect his forearm. He also uses it to transport items such as his wallet and mobile phone.

Design 1

In the first instance, the client was reviewed by our Upper Limb Review Clinic team and expressed three key functional requirements: (1) improved grip to allow bilateral function, (2) stabilisation of his flail arm and (3) reduced subluxation of his right shoulder. In discussion with the clinic team it was agreed that primary shoulder stabilisation by something more than a sling may assist with pain management through relieving traction on the damaged brachial plexus. Traditionally the Stanmore Orthosis has been utilised for people with BPI's and has delivered good results. We fitted the Stanmore Orthosis in October 2002 and this allowed the client to experience increased stability and some restoration of bimanual prehension. He quickly came to realise though that this orthosis did not meet his functional expectations because:

- It had insufficient excursion (due to the inherent design of the Stanmore Orthosis, his body shape/flexibility and limited uniscapular abduction).
- It had low excursion force translating to low prehension force

- There were difficulties maintaining the position of the terminal device during prehensile activity due to instability of both the client's anatomical shoulder joint and an inherent inability to maintain the orthosis in a forward position.

This was compounded by his high expectations of any device given his engineering background, leading to early rejection of the Stanmore Orthosis. Even though this orthosis was unsuccessful, the client's introduction to wearing an orthosis led to his determination to improve the design and he continued to work closely with the team to achieve this end.

Key orthotic design parameters for the next orthosis were:

- Increased prehension force to enable participation in engineering based tasks
- Easier operation of the terminal device – minimising the effort & excursion required
- Sufficient structural rigidity to allow the positioning of a manually locking shoulder joint to allow forward positioning (shoulder flexion) of the orthosis allowing improved positioning of the terminal device
- Simplified donning and doffing.

Design 2

In response, a second design phase was entered into. The second orthosis incorporated:

- A lightweight carbon fibre construction including a padded shoulder saddle and padded forearm section with a modified thoracic strap suspension
- An early version MICA manual locking shoulder joint
- Stanmore manual locking elbow joints
- Switch control external power system including a 9X14 switch, VASI SPM controller and a Lithium Ion power system, and
- A custom fabricated terminal device mounting for the Hosmer 5XA hook.

We felt that external power was the logical solution to the lack of body powered prehension force. The use of the Otto Bock Greifer was discussed at this stage but the client decided that the design of the 5XA hook with its fine tips was suitable for many of his workshop tasks. It was therefore decided that the 5XA hook would be linked to an external power source and control. We researched commercially available components, we sought advice from international colleagues and eventually a VASI linear actuator was fitted to a customised mounting to actuate the hook.

Again there were limitations to this design, specifically a relatively slow open & close speed. In addition, the linear actuator did not apply sufficient force to the terminal device to sustain adequate grip force on the tools and objects that the client wanted to utilise on a daily basis. While the fine tips of the 5XA were deemed useful, other inherent design issues with this terminal device include difficulty holding round objects and lack of surface area diminishing his prehensile ability. Also, the traditional positioning of the terminal device on the medial/distal surface of the forearm and the hook's fixed position restricted pronation/supination leading to significant functional limitations.

The use of the MICA shoulder joint allowed passive positioning of the terminal device but there were concerns about the longevity of the shoulder joint and its use within heavier activities. In addition, given that this shoulder unit incorporates a friction abduction joint the client's upper limb was unable to be maintained in a stable position when force was applied.

As a design revision, the 5XA hook was substituted with an OttoBock Greifer and a 9X50 linear transducer replaced the 9X14 pull switch. These changes increased prehension force and allowed proportional control, but the medial mounting position of the Greifer compromised its function. The client was also restricted in operating controls on his lathe and other machinery due to the close proximity of the hand shell.

During this period three external influences emerged and impacted upon the client's progress:

1. The client underwent unplanned shoulder stabilisation surgery,
2. He experienced marital problems culminating in separation, and
3. He began to self-mutilate his insensate hand leading to infection and subsequent phalangeal loss.

We recognised the need to take a more lateral and realistic approach to designing the orthosis. Through continued discussion with the client our revised design parameters included:

- Increased prehension force beyond what was achievable with the linear actuator,
- Increased stability of his arm to allow more accurate and stable prehension,
- A structural design that would relieve the terminal load of the relatively heavy electric terminal device,
- Better position of the electric terminal device to allow clear line of sight and space to manipulate objects with his sound left hand,
- Protection of his flail hand and arm to minimise self-mutilation,
- Given the client's ongoing cognitive issues, simplification of the cognitive capacity required to operate the prosthesis.

Design 3

These design parameters were achieved in the third design phase through:

- Deleting the shoulder girdle and shoulder joint
- Casting and moulding of the forearm/hand section closely to the body and stabilised by a thoracic strap
- Utilising the existing manual locking elbow joint
- Relocating the terminal device to the lateral aspect of the forearm (see Figure 1.)

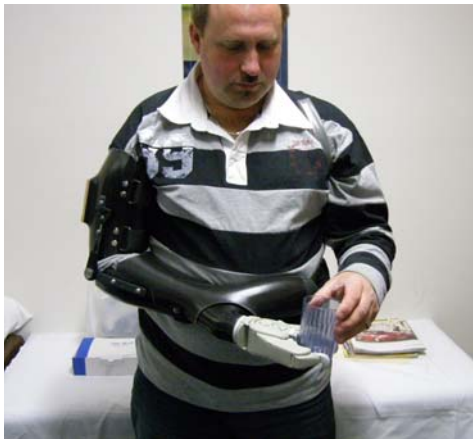

Figure 1. BPI Orthosis – Design 3

Through a trial process we found that the client preferred the terminal device to be positioned closer to his left hand. He reported that this was advantageous as it took less time to move into a bimanual prehension position.

The flexible cover over the forearm/hand section was originally designed to provide protection from welding spatter and minimise other mechanical injury. However a

secondary benefit of enclosing the forearm completely was identified. The client's access to his fingers was limited as a result of the cover and therefore opportunity for further self-mutilation was minimised.

SUMMARY

While this paper presents a single case study, many of the issues raised are common to a broad range of clients who have sustained significant brachial plexus injuries, especially those with root avulsion and associated closed head injuries. We recognise that a long recovery period frequently follows brachial plexus injury and there is often reduced motivation to utilise an orthosis. Michael and Nunely [2] state that "it is therefore imperative that the patient be actively involved in all prescription decisions from the outset; without a motivated and cooperative individual, even heroic prosthetic/orthotic interventions are doomed to failure" (p.297). Throughout this process we endeavoured to work within a client centred practice framework and, utilising the team's knowledge and experience, we were able to develop and refine the design to meet the client's primary functional need for a positive grip force on a stable base enabling maximal participation in his everyday activities.

Learning from this process

- Utilising as many commercially available components as possible leads to more predictable outcomes.
- Positioning the flail arm in alternative positions may provide a more stable base for prehension.
- Working with a client over an extended period has its own therapeutic benefits.
- Utilising a functional orthosis allows people to maintain bimanual ability and to establish a wearing pattern that may extend through to future prosthetic use.
- Current research and development into the rehabilitation and orthotic management of these patients remains limited, in particular development of more appropriate componentry appears to be a priority.

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NINETEEN YEAR FOLLOW-UP OF A BILATERAL SHOULDER DISARTICULATION (BSD) AMPUTEE

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INTRODUCTION

Management of high-level bilateral amputees poses a significant challenge for the treating parties. Selection of the most appropriate prosthetic components and controls requires knowledge of the many options available and the ability to predict which systems will most benefit the user. Long-term follow-up of these individuals is rarely reported. Through retrospective study it may be possible to identify attributes of prosthetic systems that have been successful and incorporate those attributes in future systems. This paper outlines the prosthetic management of an individual, KF, with traumatic bilateral shoulder disarticulation amputations over a nineteen-year period.

In 1986, the Prosthetic/Orthotic Clinical Service Department of the Rehabilitation Institute of Chicago (RIC) and the Rehabilitation Engineering Research Center (RERC) in Prosthetics and Orthotics of Northwestern University began a collaboration to improve the prosthetic fitting of persons with higher-level bilateral amputations. By the time of KF's admission to the RIC, a hybrid fitting concept had been developed using body-powered and electric-powered components in a complementary manner to enhance the function of the total prosthetic system [1,2]. Cable-actuated body-powered components were used on the dominant side to take advantage of the physiological proprioceptive feedback intrinsic to cable control. This prosthesis was used primarily for fine positioning and dexterous object handling and manipulation. Electric-powered components were used on the non-dominant, or assistive, side to provide higher gripping forces and greater lifting torque.

INITIAL FITTING

KF was fit with a complete body-powered system on the right side and a socket without prosthesis on the left side in October 1989. The prosthesis included a 5XA split hook (Hosmer Dorrance), a Sierra Wrist Flexion Unit (Hosmer Dorrance), a locking Rotation Wrist (USMC), an E-400 Elbow with friction turntable (Hosmer Dorrance), and a friction FAJ shoulder joint (Hosmer Dorrance). The lock control cables of the E-400 elbow and the Rotational Wrist were routed to modified Sierra Nudge Controls (Hosmer Dorrance) mounted to the socket for chin actuation. The lock control lever of the Sierra Wrist Flexion Unit was extended so that it could be actuated against KF's thigh or knee or engaged against an object in the environment.

Biscapular abduction was used to open the fingers of the split hook, rotate or flex the two wrist components, or flex the elbow using a single positioning control cable. The route of the control cable from the lateral-mounted forearm lift tab to the attachment post of the split hook passed the cable medial to the axes of rotation of the Rotational Wrist and the Sierra Wrist Flexion Unit. If the Rotational Wrist was unlocked, biscapular abduction pulled the wrist into supination. Relaxing the control cable allowed a spring to pull the wrist into pronation. If the Sierra Wrist Flexion Unit was unlocked, biscapular abduction pulled this component into flexion, and relaxing the cable allowed an elastomeric band to pull it into extension.

A pulley was set up as an excursion amplifier to double the range of cable travel. Spectra cable (TRS) and Teflon-lined housing were used to reduce friction losses.

Laminated frame type sockets with carbon fiber reinforcement were used to spread loads over a large area yet expose as much skin as possible for improved ventilation [3]. The sockets were designed to capture as much biscapular motion as possible and provide rotational stability in order to maximum function with the prostheses.

In December 1989, KF was fit with the left complementary prosthesis [4]. This prosthesis included the electric-powered Greifer (Otto Bock), the Electric Wrist Rotator (Otto Bock), the electric-powered Boston Elbow with friction turntable (Liberty Mutual), and a friction FAJ shoulder joint. The Greifer and Electric Wrist Rotator were controlled with separate chin-actuated Rocker Switches (Otto Bock) mounted to the socket. The Boston Elbow was controlled with a two-function Harness Pull Switch (Otto Bock) actuated by shoulder elevation. The frame socket on the left side had been modified with a superior cutout so that the clavicle and scapula could be raised through the opening. The switch was mounted to the anterior of the socket below the cutout with a section of harness webbing passing from the switch, over the shoulder, and anchored to the posterior of the socket.

UPGRADES AND CHANGES

No prosthesis should be considered the definitive or final prosthesis. Each major repair or replacement is an opportunity to reevaluate the client's functional goals and re-consider the design of the prosthesis in light of improved fabrication techniques and materials and new components and control schemes.

The hybrid design concept implemented in KF's 1989 prostheses proved to be both functional and versatile and was retained in each refitting. There have, however, been many changes, beginning with replacement of the right FAJ friction shoulder joint in 1991 with the, then, newly available MICA Locking Shoulder Joint.

The friction shoulder joint had been a source of frustration with the friction either too low, causing the shoulder to slip position when KF was trying to push with his prosthesis, or too high, making pre-positioning arduous. The MICA shoulder joint could be locked at 18° intervals in flexion and extension, although abduction/adduction was still friction-based. When unlocked, the joint would swing freely with trunk rotation or bending. Furthermore, the MICA joint allowed the arm to be positioned and locked over head; whereas, the FAJ joint was limited to 90° of forward flexion. A cable unlocked the MICA joint through a modified manual knee lock control (Blatchford) adapted for chin actuation. The MICA joint required more frequent maintenance than the Hosmer FAJ joint. However, the functional advantage of a locking shoulder joint was far greater than this inconvenience, and KF opted to have a locking joint retrofit to his second left prosthesis in 1992.

In 1994, a third left prosthesis was fabricated and the switch-actuated Boston Elbow was replaced with the Boston Elbow II (Liberty Technology). The control arrangement for the elbow and Greifer were also changed at that time. The earlier method of operating the Greifer with a chin-actuated Rocker Switch prevented KF from seeing the Greifer while controlling it when it was above his head. This conflict only arose as a consequence of incorporating the MICA shoulder joint. The control was changed so that shoulder elevation against the Harness Pull Switch operated the Greifer. The chin-actuated Rocker Switch, previously used for the Greifer, was replaced with two force sensitive Touch Pads for control of the elbow. The force of pressing on the Touch Pads proportionally controlled the speed of the faster Boston Elbow II.

Control of the Greifer was again changed in 1997. The Harness Pull Switch was replaced with a Linear Actuator (Liberty Technology). With this arrangement, the degree of shoulder elevation determined the speed of the Greifer or the rate at which grip force changed.

The next major change was replacement of the friction turntables (for humeral rotation) with locking HR Units (Rimjet). These devices could be locked at 22.5° intervals. Replacement was done on the right prosthesis in 1998 and on the left prosthesis in 1999. Replacement on the left prosthesis required a custom adapter to couple the HR Unit to the Boston Elbow II. The HR Units on both sides were unlocked using chin-actuated Sierra Nudge Controls.

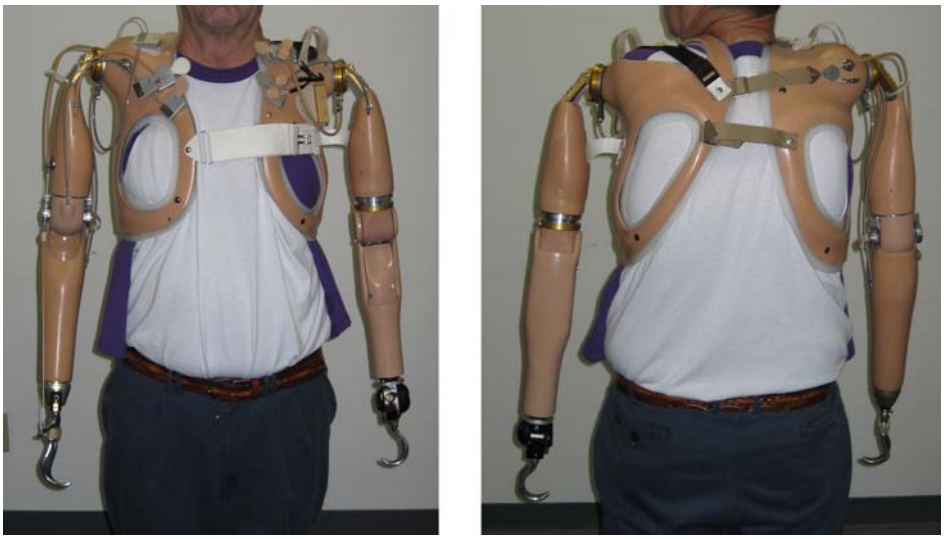
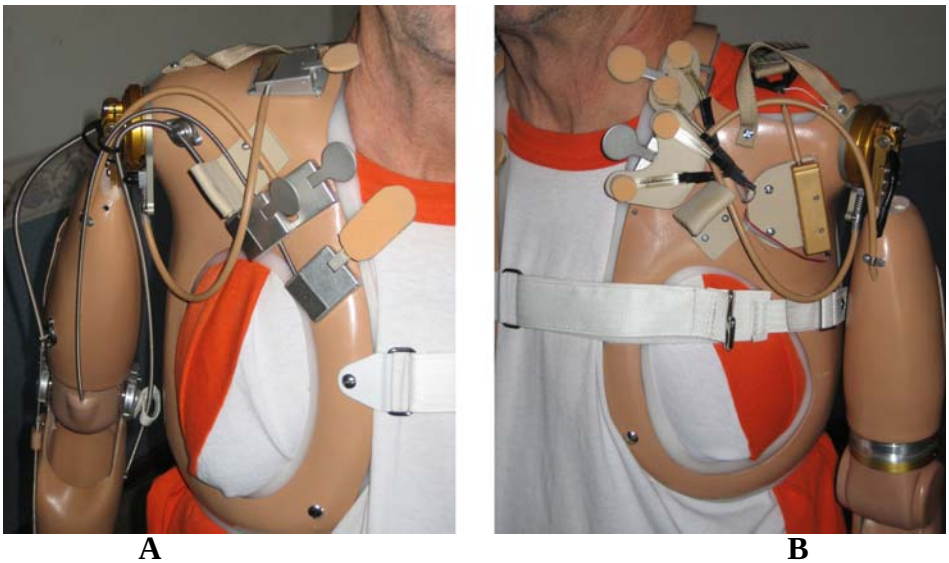


Figure 1. Anterior and posterior views of the 2006 set of hybrid bilateral prostheses.



A **B**

Figure 2. (A) Arrangement of chin-actuated Nudge Controls for the locks of components of the right prosthesis. (B) Arrangement of chin-actuated lock controls, chin-actuated Touch Pads, and shoulder-actuated Linear Actuator for components of the left prosthesis.

The most recent upgrade was done in 2006 when another set of new prostheses was fabricated (Figures 1 and 2). On the right side, the Sierra Wrist Flexion Unit and Rotational Wrist were replaced with the lighter N-Abler V Five Function Wrist (Texas Assistive Devices). Actuation of the rotation and flexion locks remained the same – chin-actuated nudge control for rotation and an extended lock release tab for flexion.

On the left side, the Boston Elbow II was replaced with the more versatile Boston Digital Arm System (Liberating Technologies). The Greifer was replaced with the ETD (Motion Control) because of KF's preference for hook-type fingers. The Electric Wrist Rotator was retained but converted to proportional control with replacement of the chin-actuated Rocker Switch by two chin-actuated Touch Pads.

Although many individual changes have been made within the hybrid prostheses, overall, the design has moved toward all locking components on both the body-powered and electric-powered sides and all proportional control on the electric-powered side. Locking components are easier to pre-position (than friction components) when unlocked and, when locked, make the prosthesis a rigid extension of the body through which high forces can be exerted on objects in the environment. Proportional control of electric-powered components, particularly higher performance components, generally improves their controllability and, consequently, their functional benefit. Reliable and repeatable control is especially important in the absence of any direct proprioceptive sense of the action of the electric component and reliance on vision.

FUNCTION AND ADLS

Within the first year after bilateral fitting KF was able to independently don and doff the prostheses and perform routine activities of daily living including; bathing, toileting (with bidet), simple meal preparation and feeding [4]. Today KF performs many tasks such as driving his truck, yard maintenance including driving a riding mower, driving a tractor, and loading and unloading the tractor on a flatbed tilting trailer. All of these tasks are performed independently on a routine basis.

CONCLUSION

Experience gained from this fitting and other high level bilateral fittings has demonstrated that a hybrid approach combining body-powered and electric prostheses has merit [5,6]. Body–powered prostheses offer proprioceptive feedback through the cable and harness and are therefore favored for fine manipulation by users of hybrid prostheses similar to those described. Electrically powered prostheses offer higher grip strength and greater live-lift capabilities and are favored for activities that require those features. The most important variable in fitting the high-level bilateral arm amputee is the user. Persons with the ability to problem solve and with the determination to master the use of their prostheses can achieve a remarkable level of independence.

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CLINICALLY PRACTICAL APPLICATIONS OF PATTERN RECOGNITION FOR MYOELECTRIC PROSTHESES

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ABSTRACT

The promise of pattern recognition for improved control of upper-extremity powered prostheses has existed for a long time. During the years of offline research and algorithm development, very little experience has been gained with real-time use in clinical and chronic settings. Our group, having the benefit of working with subjects who have undergone targeted muscle reinnervation (TMR) surgery, is at the forefront of real-world application of pattern recognition for upper extremity amputees. Based on our experiences, we highlight a progression of myoelectric control schemes from conventional control to enhanced pattern recognition control, stressing the application of simple pattern recognition schemes to replace more conventional control. These clinically practical pattern recognition systems incorporate a realistic number of electrodes and the ability to control available prosthetic components. Our experience suggests how the impending, and initial deployment of pattern recognition-controlled prostheses for daily use can be more approachable than what is depicted in high-dimension studies common in the literature today.

BACKGROUND

The use of pattern recognition for robust myoelectric control, and its promise for improved prosthesis function, has been steadily investigated since the 1960s [1-20]. Investigators around the world have developed techniques and demonstrated efficacy for systems involving anywhere from one [7, 10, 16] to hundreds [8, 15, 19] of electrodes and controlling few [2, 5, 11, 13, 18] to multiple [3, 8, 12] classes of motion. Near exhaustive studies of each algorithmic component of myoelectric pattern recognition have been and continue to be conducted, as researchers investigate the effects of different approaches to data windowing [3, 10], feature types [2-4, 10], classifier types [1, 4, 9], post-processing [3], and so on. With exceptions in most recent history [8, 20], the majority of these studies have been conducted offline, using recorded data, and have often used able-bodied research subjects as substitutes for amputees. In the meantime, clinical application of myoelectric control has persisted with ‘direct’ (or ‘conventional’) control techniques, limiting the ease with which powered prosthesis users can function. It is the authors’ view that it is a valid time for pattern recognition to be incorporated into the clinical standard for myoelectric prosthesis control. We suggest that the overwhelming amount of information and results from scientific studies of pattern recognition may be well suited to researchers but not necessarily to the clinicians who are ultimately responsible for facilitating the marriage of pattern recognition to everyday prostheses.

Direct Control

The synonymous terms ‘direct control’ and ‘conventional control’ have recently been introduced to characterize the forms of myoelectric control that do not include pattern recognition. Many configurations of direct control are clinical standards, including various strategies using the signal amplitude from one or two muscle sites (myosites) and/or coupled with switches, buttons, etc. [21]. Clinicians strive to use strategies of direct control that are

simple (to ensure user-acceptance) yet maximize function. Although these schemes are seen as reliable and robust, perhaps because they are the best we have, the function of a powered device under direct control is limited to the amount and type of control information a user can reliably present to the control system. That barrier has held R&D of device hardware to a limited pace for a number of years [22]. Recent activity in neural control mechanisms such as targeted muscle reinnervation (TMR) [23] is helping to surpass that bottleneck and fuel development of advanced powered prosthetics. Although pattern recognition has been proposed as a viable control mechanism to command these advanced (more degrees of freedom, etc.) devices, it is discussed here how pattern recognition should first benefit today's devices with a few electrodes and a small number of powered degrees of freedom.

Pattern Recognition

Proponents of pattern recognition speak of its potential for controlling multiple degrees of freedom and of how well it aligns with a future generation of powered devices. These promises are usually presented at the cost of using more myosites than the clinical norm (two for agonist/antagonist control). Many challenges exist for implementing a large number of electrodes and commercially available components for additional degrees of freedom are only beginning to become available. Microprocessor hardware capable of pattern recognition has not yet been developed onto any commercially available platforms. When these microprocessors arrive, the first true clinical implementations of pattern-recognition-controlled prostheses can be realized. It is suggested that basic setups be introduced first; that is, pattern recognition as a seamless sequential substitute for single or dual site direct control, commanding commercially available components that prosthetists are comfortable using.

PATTERN RECOGNITION: SUBSTITUTE FOR DIRECT CONTROL

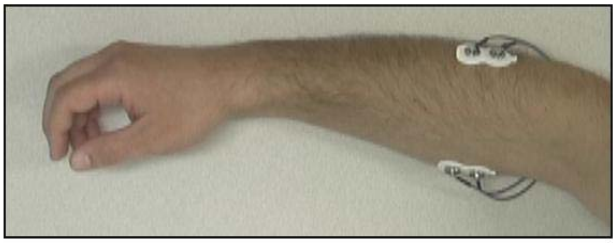
The example shown here compares a clinicians' setup of direct control to pattern recognition control using the same two antagonistic myosites, intending to operate a terminal device with powered open and close. For this demonstration, two myosites were located on an able-bodied subject by means of palpation for hand open and close (augmented with wrist extension and flexion); see Figure 1(a). In clinical practice considerable time is often also spent with an aide such as a MyoBoy[1] to help prosthetists determine ideal sites. Using custom software, ACE[2], dual-site control for hand open and close was set-up; a clinician iteratively adjusted input gains and activation thresholds for each channel while instructing the subject to open and close their hand. For the second part of the demonstration, ACE was used to separately configure the pattern recognition control for the same degree of freedom. For this, the subject was prompted to conduct two 3-second contractions of hand open and close; ACE recorded the muscle signals for these two classes along with periods of no movement, constructed an internal pattern classifier, and automatically configured the dual-site set-up. What is important to note are not the specifics of the experimental setup, rather the procedural differences when replicating a direct control set-up with pattern recognition.

[1] Otto Bock HealthCare, Vienna, Austria

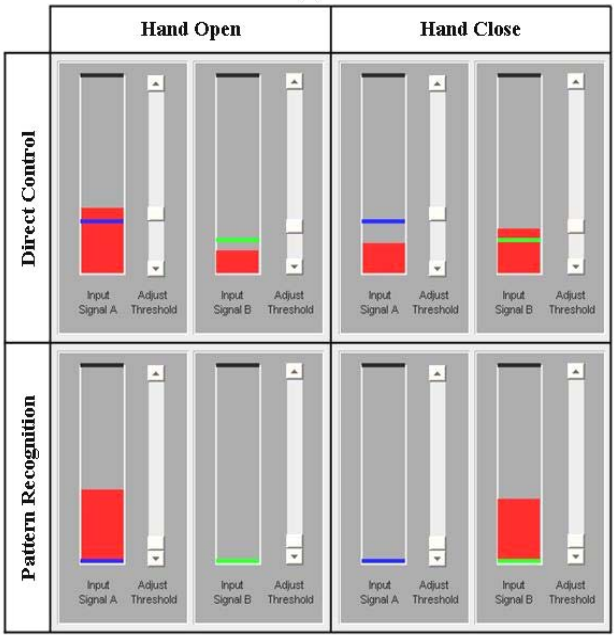
[2] Acquisition and Configuration Environment (ACE) is a clinician interface software suite developed at UNB IBME. Additional information about ACE can be found within these proceedings.

Of equal or greater importance, it can be shown that pattern recognition can provide a much 'cleaner' separation of command signals for cases of substantial myosite crosstalk (involuntary co-activation of muscles). As shown in the top row of Figure 1(b), signals from both myosites are apparent as the subject attempts to open or close the hand. This can be commonplace in current clinical practice and means the clinician must spend considerable time and effort training the user and adjusting signal gains and thresholds to achieve acceptable control. The bottom row of Figure 1(b) highlights how the signals become mutually exclusive with pattern recognition, simplifying the setup for the clinician. A short, supplemental video[3] shows the real-time direct control system (highlighting the challenge of crosstalk interference) and can be compared to the supplemental video[4] showing the real-time pattern recognition setup.

Three points can be made in support of pattern recognition substitution for direct control: (1) pattern recognition can be less susceptible to initial placement of electrodes; (2) the cumbersome (and time consuming) prosthetist process of setting gains and thresholds is replaced by a semi-automated training regimen, and, perhaps most applicable to clinicians; (3) signals affected by a large degree of crosstalk or interference can be made more distinct and useable.



(a)



(b)

Figure 1: *(a) Two myosite setup used for this demonstration; (b) Table of screen captures showing the challenge for gain and threshold selection by prosthetists using direct control (top row) and how pattern recognition can simplify setup of the same system (bottom row). (screenshots used with permission from UNB)*

PATTERN RECOGNITION: ADDING FUNCTIONALITY

Considering a prosthesis with two powered degrees of freedom (e.g. powered terminal device and wrist rotator), prosthetists must be creative with the use of external harness switches, co-contractions, fast/hard vs. slow/soft strategies, etc in order to have the device function under direct control from one or two myosites. Pattern recognition provides the potential to eliminate switching in the control system as it can decipher between the [hand open, hand close, wrist pronate, and wrist supinate] intentions of the user in a more intuitive and natural way.

Expanding on the example from above, pattern recognition was used to show enhanced functionality while using the same two myosites. The able-bodied subject performed a training session where two 3-second contractions were recorded for each of five classes: hand open, hand close, wrist flexion, wrist extension, and no motion. The complete data acquisition and classifier

[3] www.smpp.northwestern.edu/kuiken/necal_videos/2site_2motion_Direct_Control.wmv
[4] www.smpp.northwestern.edu/kuiken/necal_videos/2site_2motion_Pattern_Recognition.wmv

preparation session took approximately 1 minute. For exploration, the subject controlled a virtual prosthesis in real-time. The actual session can be seen in a supplemental video[5]. Immediately obvious is the ability to control wrist and hand functions using two myosites without the need for switching. Also, as before, clinician setup time is moved from adjustment of gains, thresholds, and switching parameters to helping guide the subject through a pattern recognition training session.

What is demonstrated here is a simple expansion of a basic case. Consider this concept a building block leading to even greater device functionality. As a prosthetist becomes more comfortable in fittings using more than one or two electrodes, pattern recognition performance will be enhanced as more myoelectric information is introduced to the system [3]. Additional myoelectric information suggests the potential to control even more degrees of freedom [8, 19].

PATTERN RECOGNITION WITH TMR SUBJECTS

Most subjects with targeted muscle reinnervation (TMR), a novel surgical technique that provides the residual muscles of amputees with neural information native to the amputated limb [23], have been fit with powered prostheses for take-home use. These semi-advanced devices have used three and four (six in one known case) myosites in place of the one or two that are customary without the surgical benefit. None of these devices employ pattern recognition; instead, prosthetists have creatively expanded on common direct control methods. Although function has improved for these shoulder disarticulation [24, 25] and transhumeral [26] amputees, the availability of pattern recognition may provide even greater benefit.

We have examined data taken from four TMR patients with the intent to demonstrate the efficacy of pattern recognition in three tasks: as a substitute of their current direct control setup, as a system controlling additional devices using the current myosites, and as a fully enhanced system controlling additional devices using additional myosites. Each subject participated in the study with informed consent approved by the Northwestern University Institutional Review Board. Electrodes were placed at predetermined locations on TMR subjects based on clinical practice (first four myosites copied their direct control prosthesis) and input from a prior study involving a high-density electrode array [8]. Using ACE, subjects were randomly prompted to hold muscle contractions for ten classes of movement for 4-seconds each, separated by 3-seconds of rest. In total, 16 seconds of data per class were recorded for pattern classifier training and 16 seconds of data were set aside for testing. Testing a classifier offline yields a measure of classification accuracy which is not a measure of, but believed to be related to, pattern recognition functionality [27, 28].

The results within Table 1 highlight a progression from direct control to enhanced control using pattern recognition. Classification accuracies in the cases where pattern recognition substitutes the direct control setup are relatively high, suggesting the subjects would function well with the motions and myosites of their take-home prosthesis and would have more intuitive control as switching or using external inputs is unnecessary. Adding four more natural motions, while maintaining direct control myosites, yields slightly lower classification accuracies that are still reasonable [3]. By adding four additional myosites, the classification accuracies return to near original levels where function was more limited. The authors advocate that all accuracies presented are clinically practical starting points for each subject and would only improve with additional subject and classifier training. The reader is encouraged to draw their own conclusions

[5] www.smpp.northwestern.edu/kuiken/necal_videos/2site_4motion_Pattern_Recognition_with_VR.wmv

on simple pattern recognition substitutions of direct control and on pattern recognition for advanced control, based on their own clinical experiences with direct control.

	% classification accuracy			
	SD-A	SD-B	TH-A	TH-B
Direct Control Setup 4 myosites: (1)EF, (2)EE, (3)HO, (4)HC (WP, WS by other†)	n/a	n/a	n/a	n/a
Pattern Recognition substituting Direct Control 4 myosites: EF, EE, WP, WS, HO, HC	94.5	97.1	87.3	91.6
Enhanced Pattern Recognition (more motions) 4 myosites: EF, EE, WP, WS, WF, WE, HO, 3grasps	84.0	80.1	72.8	80.9
Enhanced Pattern Recognition (more motions and myosites) 8 myosites: EF, EE, WP, WS, WF, WE, HO, 3grasps	95.1	88.2	85.4	89.2

†WP,WS controlled by: (SD-A) HO, HC myosites, switched by touch pad at shoulder OR co-contraction; (SD-B) 2 touch pads at shoulder, and; (TH-A and TH-B) linear transducer in harness.

Table 1: *Classification accuracies for the three pattern recognition cases and depicts the functional differences by subject. EF = elbow flexion, EE = elbow extension, WP = wrist pronation, WS = wrist supination, WF = wrist flexion, WE = wrist extension, HO = hand open, HC = hand close*

CONCLUDING REMARKS

The impending introduction of pattern recognition control to powered prostheses should not be intimidating; even if much of the published work on pattern recognition considers many myosites and many motions. In a clinically practical setup of pattern recognition, a prosthetist can benefit from 'cleaner' signals, less switching concerns, and the potential to help their patients achieve more intuitive control of more powered degrees of freedom.

The authors acknowledge Kathy Stubblefield, Robert Lipschutz, Laura Miller, Jon Sensinger, and Kevin Englehart and for providing insight into this discussion and Erik Scheme for being the pillar of software development.

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SOFTWARE UNDERACTUACTION FOR REDUCED INPUT CONTROL OF LARGE DEGREE OF FREEDOM STRUCTURES: DEVELOPMENT OF A HAND CONTROL STRATEGY FOR ADVANCED UPPER EXTREMITY PROSTHETICS

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ABSTRACT

Pattern recognition strategies have been explored for prosthetic hand control for over 20 years, but have not been commercially implemented. [1] Articulated research hands now rival human hand articulation, presenting new control challenges. Because no methods of processing signals of any source (cortical, peripheral nerve or myoelectric) are currently capable of delivering joint angle intent, the information content of user intent must be expanded. Control information for the 15 to 20 joints of a highly articulated hand must be derived from a single variable of grasp information. Two strategies are proposed for augmenting intent to control these hands. First, an array of grasp endpoints composed of the joint angles desired is created for each grasp, and the linear path to the desired endpoint constantly updated. Finger collisions are supervised by the user. Second, a larger set of grasp control parameters are explored. Using additional parameters allows more precise control of the arrangement of the fingers than whole hand grasps, a greater degree of intuitive control over the arrangement of the fingers, and the possibility of dextrous manipulation.

BACKGROUND

Grasping, Manipulation and Control

Most studies of hand use have focused on grasping: static prehensile or enclosing postures. Equally important are non-prehensile postures, as well as the movements used to generate a hand posture. Beyond strategies based on the taxonomies of grasping, truly advanced prosthetic control will depend on intuitive inputs to usefully manipulate objects. A well-designed control strategy will allow the creation of hand postures appropriate to every task, and to the manipulation of objects once in contact with the hand.

The variety of grasp taxonomies developed speaks to the complexity of simple grasping tasks, not to mention manipulation. Classifications based on purpose, which digits are used and the roles they play have been developed by Napier, Cutkosky and Kamakura. [2–5] Kamakura, Kang and Iberall further classify grasps using the non-finger contact areas of the hand. [6, 7] Such classification is interesting for prosthetic control to the extent that it could inform control inputs. In terms of effective control, it may be that the identification of the lateral pinch by name will prove less useful than the control goals for sensory inputs such as the force vectors resulting from contact, and the ability to pre-shape the hand for contact.

Postural Synergies

Neurologically inspired investigations into how hand postures are administrated during grasping have identified little difference among the discrete postures named by researchers in other disciplines. Santello *et al* found that the joint angles of the hand did not vary independently during the performance of distinct static grasps, and that two principal components can account for more than 80 per cent of the variance in joint angles. They propose that finer

thumb and finger positioning represent the less significant components that serve to further refine postures. Mason *et al* extended this work to dynamic reach to grasp, finding a higher contribution of two principal components to hand pre-shaping in reach to grasp, albeit for a limited scope of objects all representing power grasps. [8,9] Thakur *et al* performed a similar experiment, extending the interaction to manipulative exploration, finding seven principal components responsible for 90 per cent of joint angle variance. [10] These synergies were not found evident in EMG signals by Weiss and Flanders. [11] This does not mean, however, that EMG could not be used to control the principal components as method of supervisory control for prosthetic positioning.

Experimental Setup

This research is being developed in the Johns Hopkins Applied Physics Lab (JHU-APL) Virtual Integration Environment (VIE), part of the DARPA Revolutionizing Prosthetics 2009 program (RP2009). The VIE allows components of the limb control system to be quickly prototyped and tested on the modular Matlab-based platform. Input is provided through a variety of means to either the prosthetic limb or to a sensor array connected to the VIE. Thus far, skin surface electromyographical signals (EMG) have been used for all patient use. Data is collected with Duotrodes, or from a socket constructed for the P2 Intrinsic Hand, using traditional myoelectric construction with 16 pairs of LTI small dome electrodes.

Within the VIE, user interface and control algorithms can be tested. A representation of either of the arms designed by the program can be controlled in real time. Object interaction in the VIE is a planned feature, but is not yet sufficient for verification of control strategies. While prototypes of both hands have been built and demonstrated, they have not been developed into platforms that can be used to test grasping or manipulatory control. In order to verify the effectiveness of the grasp strategies designed to date, testing on articulated hand hardware needs to be done. Testing is planned using the NASA Robonaut platform at NASA Johnson Space Center in Houston, with the JHU-APL VIE system controlling the Robonaut through ethernet commands. While having three fewer degrees of freedom, the Robonaut has been under development for some time and represents a reliable test bed.

PROSTHETIC CONTROL STRATEGIES

Pattern Recognition

While intuitive user control of each joint in a hand movement is ideal, it is unlikely that it can be accomplished successfully in the near future, regardless of the signal source. Pattern recognition has been used with success in upper arm movement coordination (with the DARPA RP2009 P1 limb, for example), but extension into the hand has been more difficult. While pattern recognition for the upper arm joints provides intent data at the joint level, hand intent is currently available only as a single grasp type, classifying grasps according to the taxonomies described. These approaches have focused on classification accuracy as a measure of success rather than useability in a physical environment, and have only rarely been used to control actual hand prostheses in a clinical environment.

Multifunction Hands

Clinical use of pattern recognition within the hand has been limited to one or two grasp types, at least in part because of the mechanical capabilities of available hands. The ES and

SVEN, Swedish 'multifunction hands', in fact had two wrist degrees of freedom and only one hand degree of freedom. [12] The MARCUS multifunction hand had a precision and power grasp mechanically selected using sensors detecting object contact with the palm, but was controlled using conventional myoelectric control. [13] Farry *et al* used a pattern recognition classifier to differentiate between two grasps, controling the highly capable NASA Robonaut hand, with 12 intrinsic degrees of freedom. Once one of the grasps was identified, the robot controller would send the hand to the pre-defined position, along a trajectory called a "macro". Any change in classification engages the "open" macro. Although with more grasps and three more degrees of freedom, this is how the RP2009 P2 Intrinsic Hand was demonstrated. While the hand itself is an impressive feat of mechatronic packaging, this control strategy is almost animatronic, and is better looking than it is useful for grasping and manipulating objects. As real time virtual environments have developed, the grasp classifications have been paired with magnitude information, and been used to control animations of fixed trajectories between open and closed endpoints of a particular grasp, also referred to as macros, but different from those previously described. The user can modulate velocity with the strength of the contraction, and can stop and reverse direction. The RP2009 P1 hand, Otto Bock's Michelangelo, has a more limited grasp set, but was controlled using this approach. Changing to a new grasp involves going to a universal open position, or making an arbitrary movement to a nearby point on the new trajectory when grasps are changed. These strategies are shown in Figure 1.

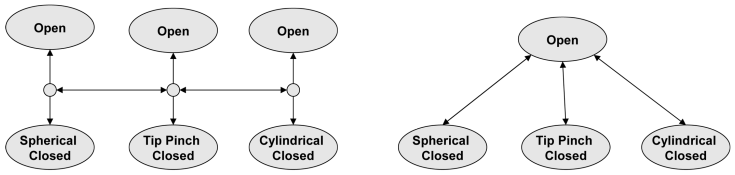

Figure 1: Traditional Macro-based Grasp Trajectories

Beyond Grasp Macros: Real-Time Shifting Endpoint Trajectories

For the greatest short term increase in prosthetic control capability, the already demonstrated capabilities of pattern recognition should be extended to the greatest effect. Because pattern recognition really represents a static endpoint goal (a closed grasp), we propose using those endpoints as control inputs for real-time arrangement of the hand. Based on a modulated macro strategy, a shifting endpoint strategy adds the capability to define new macros on the fly. The algorithm linearly interpolates the trajectory between the current hand position and the desired endpoint, whatever the state of the hand. Figure 2 illustrates an example. Beginning in an open position, the user selects a spherical grasp, and the hand begins to move toward the spherical grasp close endpoint. The figure shows that at any point along the way, the user can select another grasp, a tip pinch, for example, and the controller will interpolate a linear trajectory from the current position to the new desired endpoint. The software allows the definition of a distinct open position for each grasp, sending the hand to the particular open position appropriate for the last close command when a generic grasp open is selected. The obvious drawback of this strategy is that it is possible for the fingers to collide. Without proprioceptive feedback, the user must look at the hand to pre-

vent collisions. This is exactly the method that users of traditional myoelectric prostheses use to modulate grip strength and monitor object contact, which, although imperfect, works in practice reasonably well. Although as yet untried in hardware, this approach is promising in simulation.

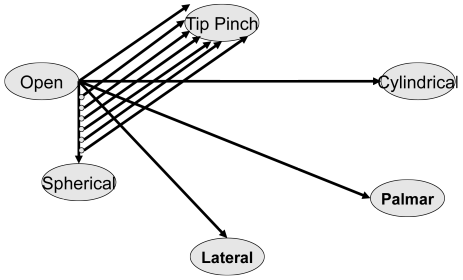

Figure 2: Real-Time Shifting Endpoint Trajectories (only some possibilities shown)

Simultaneous Endpoints: Postural Synergies as a Path to Manipulation

To extend the shifting endpoint strategy to mimic the postural synergies described by Santello and Thakur, we propose a simultaneous endpoint strategy. Simultaneous pattern recognition classifiers allow more than one degree of freedom to operate at a time, a big improvement over sequential movement where only a single degree of freedom can be operated at a time. If simultaneous classifiers are given shared control over joints, perhaps the user can usefully arrange the hand by requesting endpoints expressing multiple intent. For example, Figure 3 shows a user simultaneously requesting hand abduction and finger flexion, the combination of which could compose a spherical grasp. In this manner, the user can, rather than selecting an arbitrarily designed spherical grasp, add the appropriate amount of finger abduction, and stop, while continuing to close the hand around the desired object. As with the previous strategy, testing in actual hardware is the only way that we will know if current signal processing techniques can be married with newly capable articulated hands for truly multifunction capability.

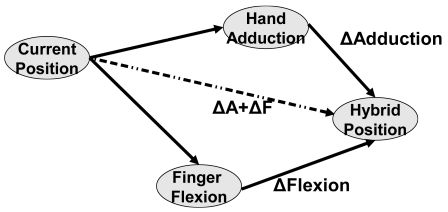

Figure 3: Simultaneous Endpoint Strategy

ACKNOWLEDGEMENTS

This research has been supported through the DARPA Revolutionizing Prosthetics Program, DARPA contract N66001-06-C8005, as well as by NSF grant DGE-0221632.

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ADAPTIVE PATTERN RECOGNITION TO ENSURE CLINICAL VIABILITY OVER TIME

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ABSTRACT

Pattern Recognition is a useful tool for deciphering movement intent from myoelectric signals. In order to be clinically viable over time, recognition paradigms must be capable of adapting with the user. Most existing paradigms are static, although two forms of adaptation have received limited attention: *Supervised* adaptation achieves high accuracy, since the intended class is known, but at the cost of repeated cumbersome training sessions. *Unsupervised* adaptation attempts to achieve high accuracy without explicitly being told the intended class, thus achieving adaptation that is invisible to the user at the cost of reduced accuracy. This paper reports a novel adaptive experiment on eight subjects that allowed a post-hoc comparison of four supervised and three unsupervised adaptation paradigms. All supervised adaptation paradigms reduced error over time by at least 23%. Most unsupervised adaptation paradigms failed to achieve statistically significant reductions in error due to the uncertainty of the correct class. One method that selected high-confidence samples showed the most practical potential, although other methods warrant future investigation outside of a laboratory setting. The ability to provide supervised adaptation should be incorporated into any clinically viable pattern recognition controller, and unsupervised adaptation should receive renewed interest in order to provide invisible adaptation.

INTRODUCTION

Myoelectric prosthesis control may be compared with recognition of a person in a photograph. If you are only concerned with the height of the person, then the silhouette of the person may be sufficient to calculate their height. Likewise, the average myoelectric signal amplitude is sufficient to calculate the speed or position of the prosthesis if some control strategy such as co-contraction will determine which joint will move.

If, on the other hand, you are asked to identify the person in the photograph, features in addition to the silhouette are required. These features may come from a variety of sources, including color (hair, eyes, skin, etc…), shape (round face, chiseled face), etc… By combining all of these features, you have a better probability of successfully identifying the person. Likewise, it is fairly straightforward to identify a desired grasp pattern or movement encoded in the electrical image produced by the muscle ensemble by looking at numerous features of myoelectric signals. Such a concept is termed pattern recognition and has been applied to myoelectric control since the 1970's [1]. Substantial progress towards a clinically viable system made in the 1990's [2] combined with the introduction of powerful microcontrollers has allowed companies to begin designing a new generation of prostheses that are capable of recognizing these myoelectric patterns, allowing for more functional and lifelike movement of multifunctional prostheses.

The above illustrations ignore an important consideration: the possibility of temporal change between training and testing. Suppose between the time you were trained to recognize a person's picture and the time you were tested, he or she cut their hair, got sunburned, and lost twenty pounds. Your ability to correctly identify this person (class) within a group of similar persons (classes) would depend on how well you had been trained, with four possible scenarios:

a) **Singular Training:** If you had only been shown the original and final image of the person, identification would be difficult.
b) **Robust features** [3]**:** If you had based your decision primarily on robust features, such as scars or tattoos, rather than on features that may frequently change, then the effect of temporal changes would not substantially affect your decision.
c) **Robust training** [4]: If during your initial training you had been shown many different images of the person (i.e., complete wardrobe, several hair cuts, pictures from different angles, etc…), it is likely that you would have recognized the person despite the changes.

This method assumes that all change falls within the permutation range of the robust training session. It also requires lengthy permutations of all of the variables that could change (i.e., losing weight and wearing heavy clothing; gaining weight and wearing light clothing, etc…)

d) **Adaptive Training** [5, 6]: If you had only been shown one image the first day of training, but you had also been shown periodic updates, your mental image of the person would adapt over time, allowing you to recognize gradual changes such as weight loss, and possibly even drastic changes if they only happened one at a time. If the identity of the person in the updated picture is provided, the adaptation is termed *supervised [6]*. If the identity of the person in the updated picture is not provided, the adaptation is termed *unsupervised [5]*.

Pattern changes over time are an unavoidable reality both in pictures and myoelectric signals. Myoelectric pattern changes can be caused by a variety of factors, including electrode conductivity changes (perspiration, humidity), electrophysiological changes (muscle fatigue, atrophy, or hypertrophy), spatial changes (electrode shifting or tissue fluid fluctuations), and user changes (adaptation or contraction intensity changes). Despite these temporal changes, the majority of reported classifiers rely on a single training session. Some studies only use signal amplitude, which is not temporally robust, in conjunction with a complex classifier [5]. Others use a slightly more robust ensemble of features with a simpler classifier [3]. Both constructions achieve high classification accuracy [7, 8], even when testing subjects with an amputation [8, 9], since they both train and test within the same session. These results may not be clinically viable, however, because they do not show robustness over time.

Other techniques seek to use robust training. Although this technique has produced encouraging results when applied to a single element of change, such as electrode shifting [4], incorporating all of the possible fluctuations into the data set requires too many permutations to be clinically viable.

The Neural Engineering Center for Artificial Limbs performs research on subjects with targeted muscle reinnervation (TMR): a novel surgical technique that provides the residual muscles of amputees with similar signal content to the normal muscles of able-bodied subjects [9-12]. Because we see the same subjects several times each year, our laboratory has turned its attention to finding a long-term clinically viable pattern recognition approach. Towards this end, we have begun a systematic analysis of the fourth solution: how to create a clinically viable classifier that gradually adapts to the user's changing patterns over time in a manner such that the user is unaware that the system is even adapting. This paper reports our progress in this critical area of clinically viable pattern recognition control.

METHODS

Five able-bodied subjects and three TMR amputee subjects (two Shoulder Disarticulation level and one Transhumeral level) participated in the study. All procedures were performed with informed consent and approved by the Northwestern University Institutional Review Board. Signals were pre-amplified and filtered using commercially available myoelectric amplifiers[1] and recorded with a custom-built 16-bit EMG amplification and acquisition system at a sampling rate of 1 kHz. Electrodes were placed equidistantly around the circumference of the proximal third of the arm with a longitudinal orientation for able-bodied subjects. Electrodes were placed at predetermined locations on subjects with TMR that had yielded the best classification accuracy using a high-density electrode array [12]. Four features[2] were extracted from each of twelve electrodes every 30 ms in 150 ms overlapped bins. A large number of classes (eleven[3]) were tested using an LDA classifier. A large number of classes were chosen to make classification difficult. Increased difficulty in turn should amplify any robustness problems, and provide room for adaptation. Custom software[4] was used to process data and

[1] Liberating Technologies, Inc. BE328 Remote AC electrodes, 30 Hz – 420 Hz -3dB bandpass filter

[2] Features included mean absolute value, # of zero-crossings, waveform length, and #of slope sign changes

[3] For amputees, the 11 classes included elbow flexion/extension, forearm pronation/supination, wrist flexion/extension, hand open, 3 self-selected grasp patterns, and no movement. For able-bodied subjects, the 11 classes included forearm pronation/supination, wrist flexion/extension, hand open, 5 grasp patterns (3-jaw chuck, lateral key, fine pinch, trigger, power), and no movement.

[4] Acquisition & Configuration Environment (ACE), a myoelectric control software program developed by the University of New Brunswick

provide a graphical user interface. Subjects alternated between training and testing trials for a total of ten pairs of training and testing data. Trials contained two repetitions of each class held for 3-4 seconds. Classifiers calculated from each training trial provided real time visual feedback for the subsequent (paired) testing trial. The experiment lasted approximately two hours, including a sum of one hour's worth of muscle contractions.

Because testing trials provided real-time feedback between each training dataset used to calculate a classifier, the static classifiers obtained over the two-hour long session may be thought of as static snapshots of an adaptive classifier. **This strategy presented a novel protocol for assessing adaptive classifiers that allowed rigorous post-hoc comparison of different adaptation paradigms while preserving the dynamic qualities of real-time adaptation.**

Post-Hoc Adaptation Comparison. Adapted classifiers were calculated post-hoc from data sets that combined data from the original classifier (of the first three classifiers, this was the classifier which had the lowest classification error) with selected samples from the testing trials. Real-time testing samples were selected based on one of two qualities, including the confidence[5] of the decision and how consistently a given class was selected. Confidence is defined as $C = \sum_{k=1}^{K} p_k \ln(p_k)$, where p_k is the probability of class k and K is the number of classes to be considered.

Two types of adaptive classifiers were compared. The first was:

Supervised Adaptation Paradigms (correct class provided):

- Supervised High confidence *(SH)*: Add samples with high confidence, with known intended class.
- Supervised Low confidence *(SL)*: Add samples with low confidence, correcting the class of low-confidence decisions.
- Supervised High/Low confidence *(SHL)*: Add samples with high or low confidence, correcting the class of low-confidence decisions.
- Supervised All *(SA)*: Add all samples, with known intended class.

The second was:

Unsupervised Adaptation Paradigms (algorithm must guess correct class):

- Unsupervised High confidence *(UH)*: Add samples with high confidence. Assume the predicted class is correct [5].
- Unsupervised Low confidence *(UL)*: Add samples with low confidence. Guess the correct class from 2nd choice guess and surrounding samples.
- Unsupervised Blip *(UB)*: Add samples that blipped to another class and then settled back to the same class. Assume the class before/after the blip is correct [13].

Data Processing. Only data following the subject's response to the visual prompt was analyzed. Tuning parameters for each adaptation paradigm were optimized based on pilot data from three other subjects. The original classifier was selected as the baseline classifier to ensure a reasonable starting trial. Adaptation paradigms added successive groups of samples to this data set for remaining trials. Relative error[6] reductions between the non-adapting classifier and the adaptive classifiers were averaged across the remaining 6-8 trials for each subject. Error from

[5] Entropy was the actual metric used, but for the purposes of this paper I have used the term confidence and adjusted the equation to better convey the meaning

[6] Relative reductions in error, rather than absolute reductions in error, are reported in the results, because relative errors retain more information content when averaged across multiple trials and multiple subjects .Information content was assessed by measuring Kurtosis, the fourth standardized moment of a distribution. Kurtosis was measured for each subject and found to be larger across subjects for relative error (2.5) than for absolute error (2.2), $p = 0.06$

the non-adapting and adaptive classifiers was fit across trials using a linear least squares regression. The statistical power of the slope determined if subject error changed over time. Minimum error was calculated by testing and training on the *same* real-time data. Any errors reported in minimum error are simply due to overlap in feature space, and may only be solved by better feature sets, more sophisticated classifiers, or better distinction by the user.

RESULTS

Adaptation paradigms frequently tagged samples to add to the baseline classifier (Figure 1). The top of this figure shows the prompted class as circles and original predictions as asterisks. The confidence of each decision is shown in the middle of the figure with a gray background. Post-hoc adaptation strategies tagged some samples to add to the training set. Adaptation strategies either used the prompted class (supervised strategies), kept the same predicted class (*UH*), or suggested a new predicted class (*UL*, *UB*) if they thought the original classifier had made an error. All of the unsupervised adaptation paradigms incorrectly tagged some samples. Non-adapting error grew larger over time for many subjects (Figure 2), and many adaptation strategies were able to reduce the average error and prevent error from increasing over time.

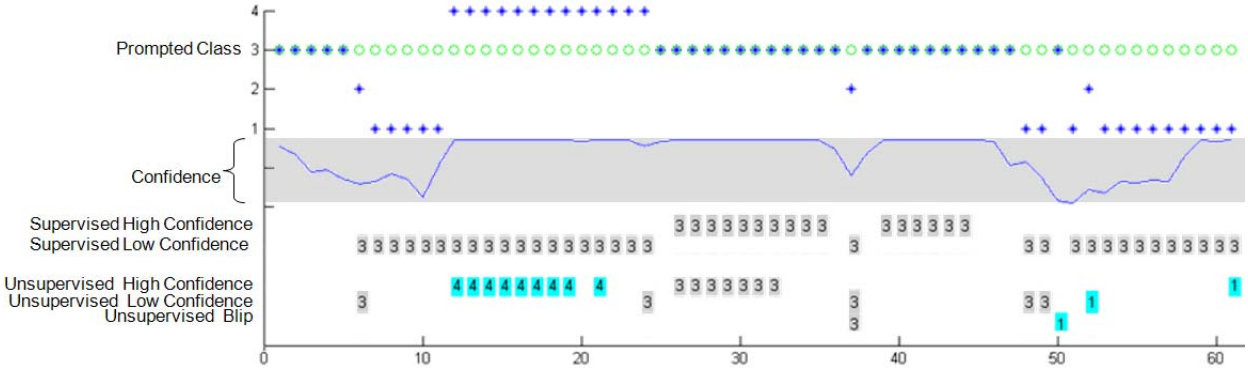

Figure 1: Composite snapshot of, from top to bottom: real-time test (circles/*), confidence of decisions (line), and samples tagged in post-hoc analysis for inclusion in the adaptive data set (numbers). Out of 11 classes, only 1 class was prompted (shown as green circles) during this snapshot and 4 classes were predicted (shown with blue *) during the real-time test. Post-hoc adaptation paradigms selected different samples to add to the adapted data set, based on criteria specific to each adaptation paradigm. Sometimes unsupervised adaptation strategies suggested the inclusion of a sample but incorrectly changed the class: these incorrectly tagged samples are highlighted in cyan.

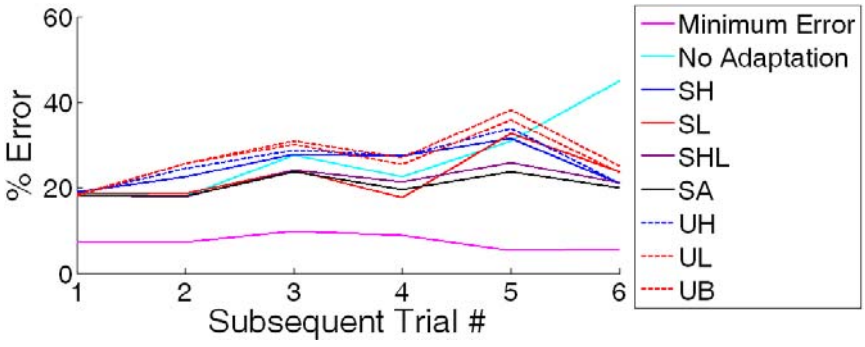

Figure 2: 2 Example of error over time for a typical subject. Subject TMR2's non-adaptive classifier produced increasing error over time.

Subjects typically had large errors when the initial non-adapting classifier was used on the data throughout the entire two-hour session. Many adaptation strategies successfully reduced this error. The average non-adapting error across subjects was 25%. All supervised adaptation paradigms reduced the error of the classifier ($p<.03$). Supervised High-confidence adaptation *(SH)* had a 27% relative reduction in error, Supervised Low-confidence adaptation *(SL)* had a 23% relative reduction in error, Supervised High/Low-confidence adaptation *(SHL)* had a 29% relative reduction in error, and adding all of the samples *(SA)* had a 30% relative reduction in error. S*A* provided more reduction in error than its more selective

alternatives (SH, SL, SHL) ($p<0.02$). None of the selective adaptation paradigms exhibited statistically significant differences from each other ($p>0.08$).

Unsupervised adaptation paradigms did not reduce error as much as supervised adaptation paradigms. The Unsupervised High-confidence *(UH)* paradigm showed the most relative reduction in error (21%), which bordered on statistical significance across subjects ($p=0.053$). Neither Unsupervised Low-confidence *(UL)* nor Unsupervised Blips *(UB)* had statistical significance ($p>0.07$) and had lower relative reductions in error (*UL*: 19%, *UB*: 16%). These results are likely due to the fact that adaptation paradigms that had to recalculate the predicted class (*UL* & *UB*) were frequently unable to correctly guess the correct class: 35% of the samples added by the *UL* paradigm and 54% of the samples added by the *UB* paradigm incorrectly reassigned the class. A small percentage (9%) of the samples added by *UH* were actually errors even though the classifier was confident they were correct.

The non-adapting classifier increased error over time for two of the TMR subjects and one able-bodied subject ($p<0.05$). Only two of the supervised adaptive classifiers increased error over time, and then only for a single subject ($p<0.05$). All of the other adaptive classifiers either reduced or maintained the same level of error over time.

DISCUSSION

On average, *Supervised High-confidence* adaptation *(SH)* added 62% of available samples, Supervised Low-confidence *(SL)* adaptation added 16% of available samples, and Supervised High/Low-confidence *(SHL)* adaptation added 75% of the available samples. *SL* provided equal reduction in error compared to *SH*, despite the fact that it only added 25% as many samples. It seems likely that, over time, repeatedly adding a large number of samples would lead to over-training, requiring a more selective inclusion criterion. *SL* likely preserves the responsiveness of the classifier to new adaptation by limiting the growth of the data set.

The Unsupervised High-confidence *(UH)* paradigm provided an implementable reduction in classification error. Although the *Supervised* Low-confidence paradigm provided a substantial increase in accuracy, the *Unsupervised* Low Confidence *(UL)* paradigm was unable to consistently predict the correct class. Future algorithms that are able to correctly identify the class in an unsupervised environment may be useful for long-term adaptation, but present implementations of *UL* are unacceptable for clinical implementation.

Some form of supervised adaptation should be incorporated into future clinically-viable algorithms. Although supervised adaptation paradigms require conscious training by the user, they are useful when the performance degrades to the point where the user is willing to push a button to go through a short training session to tune the classifier. The authors suggest that *SL* is the best supervised adaptation method, since it provides substantial reduction in error without adding a large number of samples.

Degradation Over Time. It is interesting to note that the only able-bodied subject whose non-adapting classification error increased over time was also the only able-bodied subject with previous experience controlling a pattern recognition system. All three of the TMR subjects also had extensive experience with a pattern recognition system, and two of them incurred increased error over time using the non-adapting classifier. There was no difference between experienced and inexperienced users regarding average non-adapting error, average minimum error, or average paired error, so it is unclear why experienced subjects seemed to incur more increase in error over time than inexperienced subjects. In a study that investigated the *UH* paradigm by Fukuda et al. [5], the non-adapting classifier significantly decayed over time, whereas there was no time degradation in the present study for many of the able-bodied subjects. Both studies used a single session that extended 1.5 – 2 hours. Fukuda et al. only used a remapped version of signal amplitude, which may be sensitive to fatigue. Classification of an ensemble of time-domain features has been shown to be robust to fatigue [3]. As a result, the ensemble of features used in this study may have prevented a need for fatigue-based adaptation due to the more robust set of selected features.

A linear discriminate analysis (LDA) classifier was chosen for this study, rather than a neural network as used in other adaptation studies [5]. Previous studies have shown that LDAs perform just as well as neural networks for static systems [7]. Although it is conceivable that a nonlinear classifier may have adapted better than an LDA, low levels of minimum error for each subject suggest that the LDA provided sufficient room for adaptation.

This study only involved a single two-hour session per subject. Studies that investigate adaptation over days or months require take-home prostheses capable of myoelectric pattern recognition. Once these prostheses are available, more practical results will be obtainable. The ability to design and evaluate adaptation algorithms is limited in part by our field's conceptual weakness regarding what constitutes robustness or optimal performance. Adaptation work would greatly benefit from future research that provides a better mathematical and therapeutic framework through which to understand these critical concepts.

CONCLUSION

All supervised adaptation paradigms provided reduced classification error of a myoelectric system. Incorrect classification prevented unsupervised paradigms from achieving significant results, with the exception of a high-confidence unsupervised paradigm.

Acknowledgements: The authors thank Aimee E. Schultz, Kathy Stubblefield, and Laura Miller for reviewing this paper. This work was supported by the NIH National Institute of Child and Human Development (grant no. R01 HD043137).

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SURFACE EMG CLASSIFICATION USING MOVING APPROXIMATE ENTROPY AND FUZZY LOGIC FOR PROSTHESIS CONTROL

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1.0 ABSTRACT

Electromyographic control systems based on pattern recognition have become an established technique in upper limb prosthetic control application. This paper describes a use of fuzzy logic to discriminate different hand grip postures by processing the surface EMG from wrist muscles. A moving data window of two hundred values is applied to the SEMG data and a new method called moving approximate entropy is used to extract information from the signals. The analyses show differences at three states of contraction (start, middle and end) where significant dips can be observed at the start and end of a muscle contraction. Mean absolute value (MAV) and kurtosis are also used in the extraction process to increase the performance of the system. The extracted features are fed to a fuzzy logic system to be classified and select the output appropriately. The preliminary experimental result demonstrates the ability of the system to classify the features related to different grip postures.

2.0 INTRODUCTION

Surface electromyography (SEMG) is used to evaluate and record the electrical activity of skeletal muscles. It is a non-invasive technique and has been applied to wide range of applications. Generally, it has been used in clinical neurological problems and now has become an established technique in prosthetic control application that is based on pattern recognition [1].

An electromyographic control system (ECS) is based on pattern recognition system (ECS) using pattern information extracted from the analysis of the SEMG signals and determines the control signal that will select the final output of the device operation. For example, Englehart and Hudgins used the pattern recognition ECS to investigate four channels of upper limbs' SEMG signal to discriminate multiple classes of hand and wrist movement [2].

This paper describes the development of a control algorithm based on pattern recognition for prosthetic hand application. The objective is to design a system that should be able to select four different hand grip postures (e.g. spherical, lateral, tripod and power) by using two SEMG signals (wrist flexor and extensor muscles) as the control channel. A new approach called moving approximate entropy (mApEn) is the main method used to extract information from the SEMG signals with a fuzzy logic to classify the information.

3.0 EXPERIMENTS

3.1 Data Collection

SEMG data were recorded from twenty normal healthy volunteers' wrist muscles. Surface electrodes were placed on the flexor carpi ulnaris (FCU) and extensor carpi radialis (ECR) at the forearm with a reference electrode at the elbow. The volunteers were asked to do three different tasks related to the selected muscles which were wrist flexion/ extension, co-contraction and

isometric contraction. Twelve contractions of ten seconds were recorded from each subject. Detail description of the procedures has been reported in a previous study [3].

3.2 Signal Processing

The recorded SEMG data was post processed where the whole data were divided into overlapping segment with a length 200 samples (130 ms duration) with a delay of one point to the next segment. Raw SEMG data was used and with this segmentation technique, the information in the signal could be preserved.

For feature extraction, approximate entropy (ApEn) is used as the main method. ApEn is a method introduced by Pincus [4] to investigate the regularity/irregularity of a signal. It measures the repetition between two vectors *m* and *m+1* within a tolerance range of *r* of the standard deviation of the data. With *m=2, r =0.2*; ApEn is obtained from the following equation:

$$ApEn(m,r,N) = \frac{1}{N-m+1}\sum_{i=1}^{N-m+1} \ln C_r^{\,m}(i) \; - \; \frac{1}{N-m}\sum_{i=1}^{N-m} \ln C_r^{\,m+1}(i) \qquad (1)$$

where N is the number of samples. The ApEn calculation returns a nonnegative number where higher value shows irregularity of the data and more regularity when lower value is obtained.

Several other methods (mean absolute value (MAV), number of zero crossings, standard deviation, kurtosis and skewness) that had been reported in the literature [5, 6] had been investigated and only MAV and Kurtosis showed useful information. From the assessment, MAV and Kurtosis were selected to be used in the feature extraction process to increase the performance of the developed system.

A moving data window was applied to the data sequence and three features (ApEn, MAV and Kurtosis) were extracted within the data is calculated repeatedly. Moving features were obtained as the windows moves point by point along the time axis. From visual inspection of the analysis from 20 subjects, a significant pattern from the three features can be observed at different states of contraction, namely start, middle and end of a contraction and detail descriptions can be referred to previous study [3].

The average at each state from each feature was calculated. Table 1 shows the summary of the analysis for average ApEn, average MAV and average kurtosis at three different states of a contraction: start, middle and end, also when the muscle is relaxed. The data from Table 1 is used as the placement of the fuzzy sets, discussed in the classification section

TABLE 1: Summary of the feature extraction process from FCU and ECR muscles

	STATE OF CONTRACTIONS							
	START		MIDDLE		END		RELAXED	
	FCU	ECR	FCU	ECR	FCU	ECR	FCU	ECR
ApEn	0.45	0.5	0.75	0.75	0.4	0.45	0.9	0.85
MAV	10	40	50	110	20	30	5	10
Kurtosis	7	5	3	4	10	15	3	3.5

3.3 Classification

A classifier's function should be able to map different patterns and match them appropriately, in this case, able to select different hand grip postures. The extracted features were then fed into the fuzzy logic (FL) classifier for the developed control system. FL developed by Lofty Zadeh [7]

provides a simple way to arrive at a definite conclusion just based upon imprecise input information.

For the FL classification analysis, triangular shape of the membership function (MF) for the inputs (Table 1) and output, and centroid method for the defuzzification are used. The rules are created based on the information from the states of contraction. For initial testing, the control system is tested with the recorded SEMG signal and implemented using MATLAB/Simulink. Figure 3 shows the flow diagram of the control system.

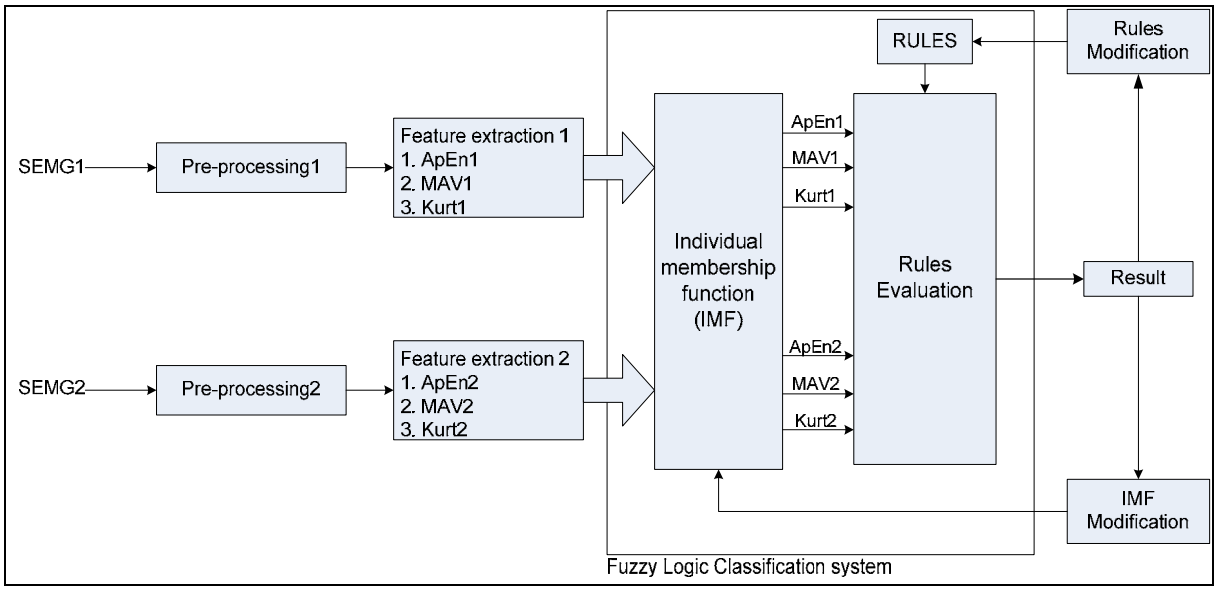

Figure 3: The flow diagram of the control system with FL classification components

Figure 4 shows the result of the FL classification process from two SEMG signals (FCR and ECR) during wrist flexion (FCR active, ECR inactive). Two examples are given and each took about 650ms duration. For preliminary testing, the four hand grip postures are represented by STATE (S1, S2, S3 and S4) and its MF is shown in Fig. 4C. The final classification result shows that the system is able to select S1 to S3 according to the features.

4.0 CONCLUSION

With moving ApEn as the main method in the feature extraction process, the preliminary result obtained demonstrates the ability of the system to classify the features related to different grip postures. In the near future, further assessment of systems' performance will be continued and the development of the control system for real time application will be carried out.

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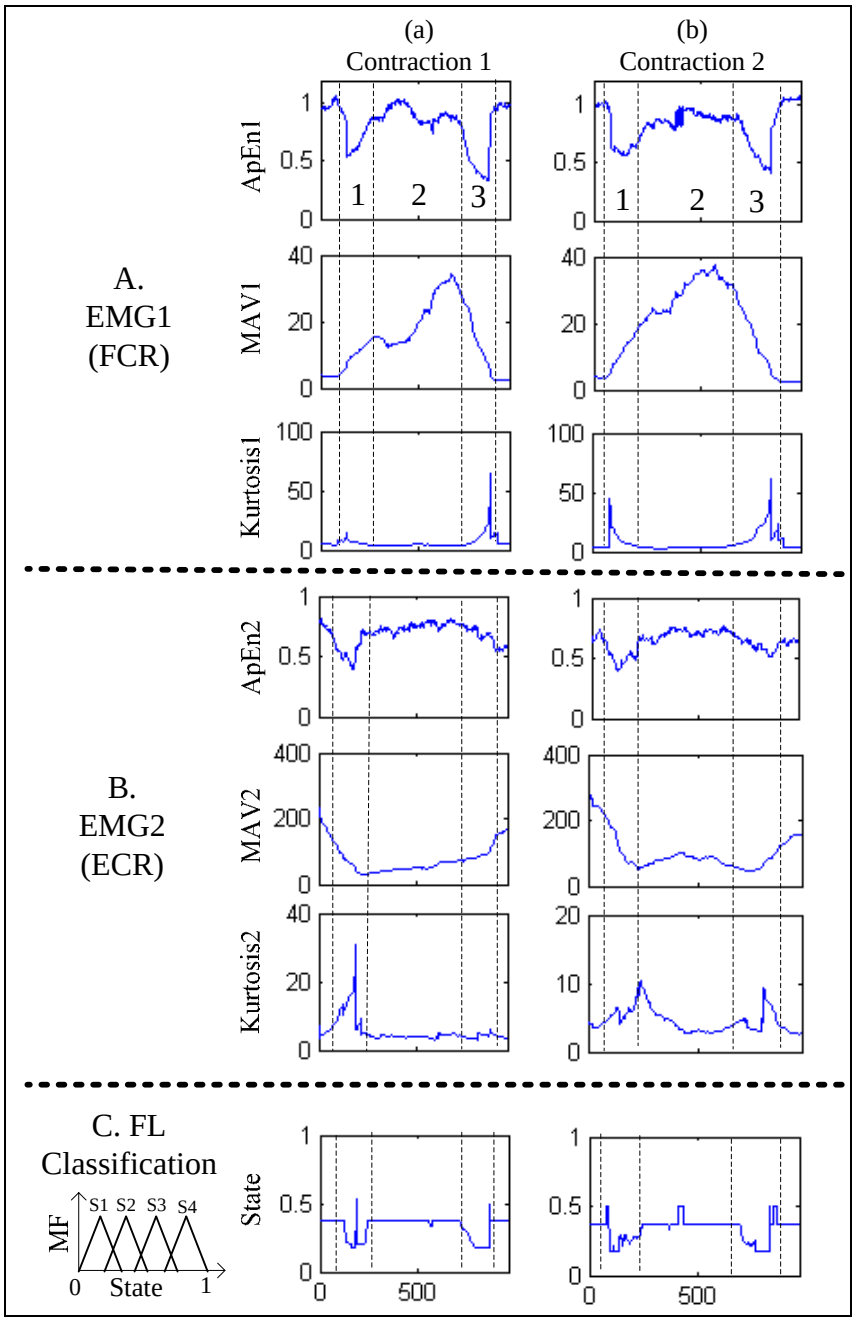

Figure 4: The extraction features from FCR (A) and ECR (B). (C) is the result of the classification based on the MF (S1: 0 – 0.35 ; S2: 0.2 – 0.55; S3: 0.4 – 0.8; S4: 0.65 – 1.0) assigned for the output. 1, 2 and 3 in the plots indicate start, middle and end of contraction respectively.

MULTIFUNCTION PROSTHESIS CONTROL USING **I**MPLANTED **M**YO**E**LECTRIC **S**ENSORS (IMES)

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I. INTRODUCTION

Persons with recent hand amputations expect modern hand prostheses to function like intact hands. Current state-of-the-art electric prosthetic hands are generally single degree-of-freedom (opening and closing) devices that are controlled using only two muscle signals. As a result, most state-of-the-art devices fail to meet user's expectations and tend to be under-utilized or rejected. [2]. In this paper we describe the development of implantable myoelectric sensors (IMES) that will allow us to record myoelectric signals from up to 32 muscle sites. Most of the eighteen extrinsic muscles of the hand remain intact following hand amputation. The goal of this work is to develop the means to control for a multi-degree-of-freedom prosthetic hand that is capable of true dexterous manipulation. The development of IMES allows us to create many more control sources than has been possible in the past, greatly increasing the number of degrees-of-freedom we can control in a prosthetic system.

II. TESTING

Testing was conducted at a number of levels to both validate the design specifications and to demonstrate the ability of the IMES system to measure muscle activity (EMG signals) *in vivo.* A progression of experiments - bench tests, in vitro tests, acute in vivo experiments, and chronic *in vivo* experiments provided this validation and demonstration of the IMES capabilities. The bench tests were performed to validate the various IMES system components and design specifications. An *in vitro* experiment demonstrated that the magnetic and radio frequency link functioned correctly when an IMES was placed in muscle tissue, a necessary precursor to implanting IMES in cats. The animal experiments consisted of both a series of acute *in vivo* experiments to show the ability of the IMES system to accurately acquire physiologically generated EMG, and a series of chronic *in vivo* experiments show the ability of the IMES to provide live EMG data reliably over time. In all experiments the IMES implants are powered magnetically via the inductive link and telemetry is acquired via the reverse telemetry bands.

Protocols

1) **Validation.**

An IMES system consisting of a Telemetry Controller, integrated magnetic drive with RF receiving antenna coil, and an IMES implant was validated on the bench against design specifications. A set of sine wave test signals, varying logarithmically from 1 Hz to 10 KHz, was used to characterize the system. IMES data was recorded via a custom LabVIEW 7.1 (National Instruments, TX) virtual instrument configured to interface with the Sigenics IMES telemetry server. The test signal set was created using a LabVIEW 7.1 virtual instrument and output via a National Instruments NI-6221 A/D board. Samples were generated at a minimum rate of 50 samples per epoch

2) **System Comparison.**

The IMES system was compared to a Noraxon TeleMyo 2400 (Noraxon, AZ) wireless EMG system. The Noraxon TeleMyo is a commercially available sixteen channel EMG acquisition system. The system response of the Noraxon system was determined by passing a sine wave test signal set through one channel of the TeleMyo 2400. The gain, high and low filter cut off frequencies were determined as above. Using the system response characteristics of both systems, a set of setup parameters was chosen to most closely match the setup for the two systems. This was done because we were interested in determining how closely the IMES response would match that of the Noraxon System for a given set of setup parameters. Both

systems then recorded an alternating series of positive and negative step functions n=100 occurring at 1Hz to provide an estimate of the step response of both systems**.** The low frequency of the step function allowed both EMG recording systems to return to steady state after being perturbed.

3) **In Vitro**

Once the bench tests were completed and as a precursor to testing the IMES system with live animals the IMES system was evaluated in an *in vitro* model. This in vitro model consisted of a shank of lamb complete with bone. A shank of lamb was chosen because it has muscle tissue surrounding a longitudinal long bone(s), and cross-sectional area similar to that of a human forearm. A 15 mm long incision was made 12 mm deep oriented parallel to the long bone of the lamb shank. An IMES implant was placed into the incision and sutured closed. A stimulating monopole electrode was placed 5 mm from each of the IMES endcaps, to the same depth as the IMES, oriented along the axis of the implant. A pair of fine wire electrodes were implanted to the same depth as the IMES directly beside the implant, and connected to the Noraxon system for system comparison. A series of individual short duration (100-1000uS) monophasic stimulus pulses were input across the monopole electrode pair and recorded by the EMG recording devices.

4) In Vivo (acute)

Performance of the IMES *in vivo* was evaluated in both an acute and chronic animal preparation. The goal of the acute experiments was to investigate how well the IMES system measures a natural EMG signal, which was elicited with the cross-extension reflex [1] in a decerebrated cat preparation. The crossed extension reflex is by definition a natural form of activation, producing normal recruitment and rate modulation [3]**.** Cats were chosen because the calf muscles of the cat are similar in size and orientation to the small muscles of the human forearm [3]**.** In this experiment three IMES implants were implanted into the ankle extensor muscle group in the cat (calf muscles); Lateral Gastrocnemius (LG), Medial Gastrocnemius (MG), and Soleus (S).

The deeply anesthetized (1-3% isoflurane) animals (n = 3) were mounted to a rigid stereotaxic frame (*Kopf Instruments*). The extensor muscle group was surgically exposed, leaving all major nervous and vascular structures intact. A cuff electrode was placed around the tibial nerve before the branching plexus of the nerve to form the medial gastrocnemius and lateral gastrocnemius-soleus nerves. Each IMES was implanted by means of a surgical cut-down into the belly of the target muscle with the long axis of the IMES oriented parallel to the muscle fibers. This orientation was chosen to minimize the predicted pick-up volume of the IMES [4]. A set of bipolar fine wire electrodes were implanted into each of the muscles, parallel to and to the same depth as the IMES implant for use with the Noraxon system. The skin was stapled shut over the surgical site to prevent dehydration of the tissues. The Power/Telemetry coil was placed around the hind limb, and oriented concentric with the implant location; the implant(s) were powered and tested to confirm they were intact. The animal was decerebrated as per [5]**.** Gaseous anesthesia was then discontinued, and the animal was allowed to breathe room air.

Direct stimulation EMG was elicited by stimulation of the nerve directly via the nerve cuff using a Grass stimulator (PSIU6) (200uS pulse width, 10Hz, 10 pulses). Stimulation consisted of several trains of monophasic pulses with stimulus intensity ranging from threshold to four times threshold, as determined for each animal. Separate recordings were made for each level of stimulus intensity. EMG was then elicited via the crossed extension reflex. The crossed extension reflex was activated by administering painful stimuli to the contra-lateral hind paw of the animal or by direct electrical stimulation consisting of high-frequency monophasic pulses to

the tibial nerve of the contra-lateral hind limb. Multiple crossed extension events were recorded for each animal.

Cross talk and IMES field sensititivity were evaluated by severing the MG branch of the tibial nerve which was identified previously. Direct stimulation and crossed-extension elicited EMG were both acquired again, per above. Severing the MG branch of the tibial nerve eliminates the myoelectric activity of the MG.

5) In Vivo (chronic)

Evaluation of chronic IMES system function was performed by inserting IMES implant(s) into the Tibialis Anterior (TA) and lateral gastrocnemius (LG) of three cats and allowing the implantation site to completely heal. The TA and LG are agonist-antagonistic lower hindlimb muscles (i.e., ankle flexor vs. ankle extensor); as such they tend to be excited out of phase with each other during normal walking – allowing us to measure two independent signals and at the same time to see the level of cross talk the IMES would pick up between these two muscles.

Implantation was performed in a sterile surgical suite under a general anesthetic (isoflurane). A small incision was made in the lower hind limb, and the implants were placed into the muscle tissue via small incisions and sutured closed as in the acute procedure, but with absorbable suture (4.0 Vicryl). The IMES were powered and ordered to transmit their numeric identifier to confirm function. The skin incision was then closed, and the animal allowed to heal. A custom jacket (*Harvard Scientific)* was modified with a set of elastic straps to contain a small, silicon encased power/telemetry coil. The cats are acclimated to the jacket and coil apparatus for a period of two weeks prior to implantation, and re-acclimated for a minimum of two days post recovery. EMG was acquired during natural walking, encouraged by enrichment toys and food rewards. Data was acquired at 6050 Sample/second per implant. IMES internal filter corner frequencies are set at 4Hz and 6,600Hz respectively.

III. RESULTS AND DISCUSSION

a) General Notes

A time synchronization algorithm was used to calculate the normalized continuous RMS voltage levels for both records and make use of the maximum cross-correlation point to allow for time synchronization. All signals are post processed with a 25Hz-1000Hz 3rd order Butterworth band pass filter to eliminate low frequency motion artifacts in the Noraxon data channel. All EMG data is normalized to allow for processing and comparison. Cross correlation and MSC were used as measures of signal similarity.

b) Validation and System Comparison.

High and low corner frequency fidelity was determined by plotting signal amplitude against frequency and calculating frequencies at which the amplitude of the measured signal dropped to -3dB of the maximum measured signal amplitude. System parameters of the TeleMyo 2400 system were calculated in the same manner. To determine the step response of both systems, a 50 second segment of each record was divided into five second windows for averaging purposes. The mean n=10 of the maximum cross correlation values is 0.850. This degree of cross-correlation shows that the IMES system will produce responses similar to that of an EMG system which is currently in clinical use, suggesting that the IMES system is able to measure EMG accurately.

c) In Vitro.

The mean n=5 of the maximum cross correlation values is 0.767 which indicates that there may be a small degree difference in detected signals. This may be associated with the conformation of the IMES device endcaps vs. the fine wire recording electrodes; producing different spatial filtering properties [6]. This shows a high degree of coherence at low frequencies, with the correlation in frequency falling off as the frequency increases, which would be expected with varying spatial filters [8].

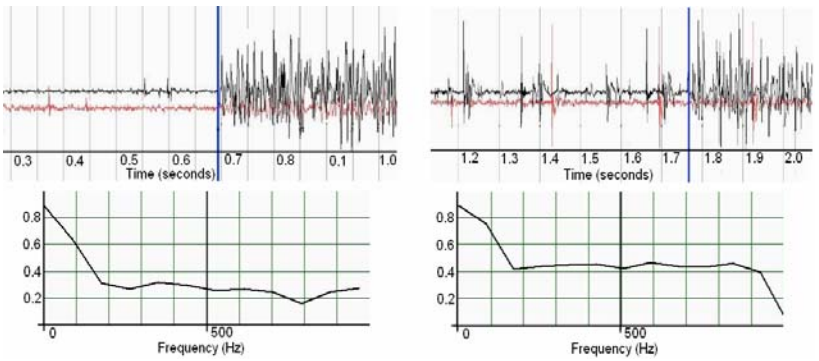

Fig. 2: Left: Intact medial gastrocnemius nerve. Right: medial gastrocnemius de-enervated. Both graphs show onset of crossed extension reflex. Normalized (voltage) raw EMG (top) black is soleus, red is medial gastrocnemius. Vertical bar indicates reflex onset. Magnitude

d) In Vivo (acute)

One second of each reflex event containing both pre- and post-onset EMG data was processed. Maximum cross correlation between signals does not exceed 0.080 which is to be expected from devices which are measuring composite stochastic signals from multiple sources. Evaluation of the pick up field of the IMES was accomplished by comparing the RMS voltage of the detected signal in the medial gastrocnemius before and after the medial gastrocnemius was denervated. Average RMS voltage measured in the medial gastrocnemius during a crossed reflex extension decreases to less than 5% of the RMS voltage measured in the Soleus (average of 27mm separation between implants) during the same reflex event. This corresponds to the levels predicted by Lowery *et al* [4]. One can also see (Fig. 2.) the increase in MSC of the detected IMES signals indicating that cross talk now accounts for a larger percentage of the total signal detected in the medial gastrocnemius. Decrease in the RMS of the signal coupled with an increase in coherence **(see Fig. 2.)** suggest the detected signal is composed of cross-talk from the adjacent muscles while at the same time being within the absolute limits predicted by Lowery *et al* [6].

e) In Vivo (chronic)

One second of data chosen at random during the course of normal walking was used for comparison**.** Both signals were normalized and post processed with a 25Hz-1000Hz 3rd order Butterworth bandpass filter to remove motion artifacts. The acquired EMG from the chronic *in vivo* experiments shows a maximum cross correlation of 0.09. MSC between the two measured signals does not exceed 0.40 and for most of the recording spectrum coherence is lower than 0.10**.** These coherence and cross correlation values show a distinct independence between signals attesting to the ability of the IMES system to make focal EMG measurements from multiple muscles in close proximity without picking up excessive cross-talk.

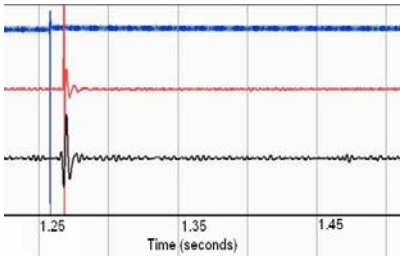

Fig.1. Characteristic normalized (voltage) impulse response of both systems through tissue. Timescale in seconds. Black trace, IMES 2520 sps. Red trace Noraxon 3000 sps. Blue trace input signal 3000 sps.

f) General Discussion

During the course of these experiments a number of issues had to be addressed. The nature of the USB interface necessitated the development of signal processing techniques which allow accurate comparison of signals acquired from disparate sources. This synchronization relies on the presence of distinct time stamps in both data streams. These time stamps were either impulse or step responses in the bench top experiments, or single motor unit action potentials in the acute cat experiments. It is important to remember that this synchronization is only necessary when trying to correlate signals acquired by the IMES with signals from a 2nd data acquisition system; data acquired from multiple IMES remains synchronous. Another issue that had to be addressed was due to the nature of the phase locked loop in the IMES implants. The resonant frequency of the class-E oscillator can be influenced by the presence of ferrous materials in or near the magnetic field. This frequency shift propagates through all aspects of the IMES system operation.

To address the effect the frequency shift has on sampling rate; all recorded signals were resampled to the nominal sampling frequency via spline interpolation. The frequency shift also affects the nominal reverse telemetry frequency, and can increase the amount of bit error the system sees due to shifting the reverse telemetry frequency outside of the range of frequencies the antenna and associated circuitry is designed to accommodate. This has been partially addressed by altering the filter corner frequencies on the receiving antenna circuitry to accommodate a larger range of frequencies. A thorough examination of the ASIC design revealed a pair of positive temperature biased transistors in the phase loop circuitry which were out of specifications, leading to larger than predicted frequency shifts when the implants were operating at body temperature. To address this issue, the nominal operating frequency of the IMES system was decreased to compensate for the slight increase in the carrier of the reverse telemetry signal. This has been remedied in the newest revision of the implant ASIC; which uses pair of matched positive and negative temperature biased transistors to maintain a reliable telemetry frequency independent of temperature.

2) Ongoing work

We are continuing to monitor the three chronic animals for signs of implant migration and to provide data for the analysis of data over time, with which we plan to address the consequences of device encapsulation on both data and signal quality. To date, we have not seen signs of implant migration or a decrease in the quality of telemetry signal and have not seen a qualitative decrease in the EMG data but are awaiting the conclusion of the study period to perform a quantitative analysis of the chronic EMG data. We are additionally in the process of performing an in depth quantitative analysis of all of our data in order to address the reliability and consistency in regards to the data acquisition abilities of the IMES system.

IV. CONCLUSION

The IMES system is capable of measuring focal intramuscular EMG comparable in both the time and frequency domain to commercially available clinical EMG systems. The use of implantable sensors in place of percutaneous wires makes the IMES system a reliable and robust platform for any EMG measurement application where a coil, flat or circular, can be accommodated on the body, such as a flat coil over the pectoralis or a cylindrical coil around a residual forearm. The IMES system is not limited to upper-limb prosthesis control and has application in lower-limb prosthetics as more powered components enter that field. In addition, IMES systems have application in experimental research where intramuscular recordings need to be made over long periods of time [7]. Using IMES obviates the need for percutaneous wires,

and can be viewed as a platform technology for making long-term intramuscular recordings. We have demonstrated the functionality and reliability of the system on the bench and we have fully operational systems that have been tested both acutely and chronically in cats. Six out of six chronic implants are completely operational 10 months after implantation. Clinical experience with implantation indicates minimal difficulty in implantation and minimal discomfort. Future research will include evaluation of the IMES system for multifunction prosthesis control.

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CANCELLATION OF FORCE INDUCED ARTIFACTS IN SURFACE EMG USING FSR MEASUREMENTS

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INTRODUCTION

As multifunction prostheses become increasingly common, there is a need for improved control signal quality in order to control all the functions. Most signals commonly used for prosthesis control are sensitive to sweat, motion and external forces [1], which impairs prosthesis control performance.

We have developed a prototype surface electromyogram (sEMG) sensor with three built-in force sensing resistors (FSRs) for measuring the external forces, which may be used to cancel artifacts caused by these forces. The performance of the sensor as an estimator of muscle force is presented in this paper. The sEMG and FSR signals have also been tested individually, as a reference for the performance using the combination of these signals.

MATERIALS AND METHODS

The sEMG sensor unit was built from the metal electrodes of an Otto Bock 13E125 device, mounted with the original spacing and wired to an external preamplifier.

FSRs were chosen for force sensors due to their flatness and simplicity of use. Three individual FSRs allow both magnitude and position/direction of an external force to be estimated, factors both of which may be relevant for the artifact identification. Initial tests used an FSR component that was readily available. It is anticipated that with more appropriately sized sensors, the entire device will fit into a prosthesis socket. The sensors were sandwiched between two layers of acrylic glass using soft double sided tape (Fig. 1). The electrodes were attached to this structure with the reference electrode at the centre of the FSR array.

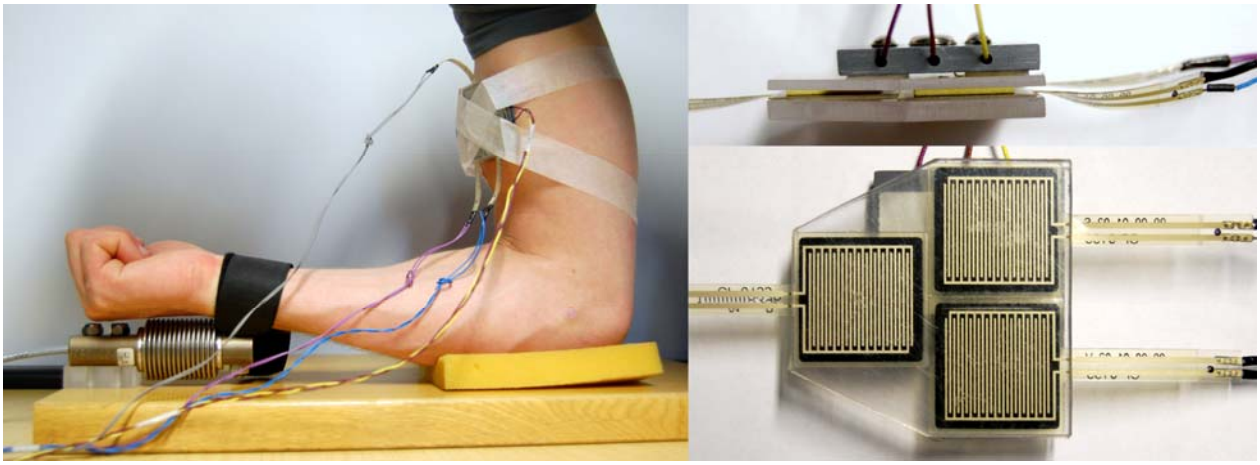

Fig. 1: Experiment setup and a close view of the sensor unit.

The device was taped to the *m. biceps brachii* of a healthy subject and tested by simultaneously measuring sEMG and FSR outputs while muscle contraction force was measured using a load cell (Fig. 1). The sEMG signal was pre-processed with a non-linear myoprocessor described in [2]. External forces in random directions were applied to the sensor during the

measurements in order to induce artifacts. Data was collected at 218 Hz for approx. 50 s. Three data sets were acquired; a *training set* and a *validation set* collected immediately after each other, and a *test set* acquired after having removed and then reapplied the device to the subject's arm.

Multilayer perceptron (MLP) networks with different numbers of hidden nodes (2-25 nodes, 10 MLP networks of each size) were employed to estimate the muscle force based on sEMG and FSR signals. Following MLP training and validation, the best 50% of the MLP networks of each size were chosen for final assessment using the test set. A linear and a quadratic mapping function were also fitted to the training set for comparison.

RESULTS

Fig. 2 presents an example data set with all recorded data. Note the two central peaks in the FSR signals, which are not accompanied by peaks in the load cell signal; these represent artifacts. The result of the force estimation, using the test set and an MLP network and a linear mapping function, respectively, is presented in Fig. 3.

Fig. 4 shows the estimated against measured force for the test set after training and validating the MLP network. Note the presense of hysteresis in the FSR based estimate and the apparent treshold levels in the sEMG based estimates. Also note the presence of force artifacts in both sEMG based graphs, evident as significant force estimate values at approximate zero load cell force.

The root mean sqare error (RMSE) rates for the different combinations of sEMG and FSR as inputs are presented in Fig. 5. No reduction in RMSE was detected when increasing the number of hidden MLP nodes beyond n=4.

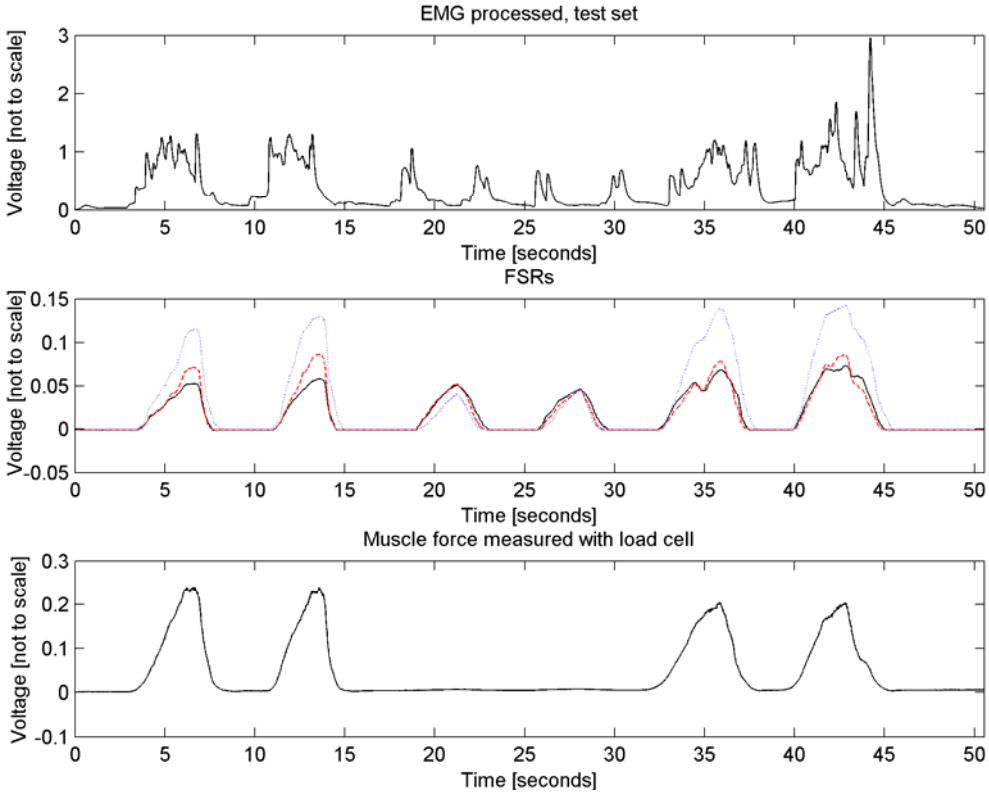

Fig. 2: Test set containing measurements from sEMG, 3 FSRs and the load cell.

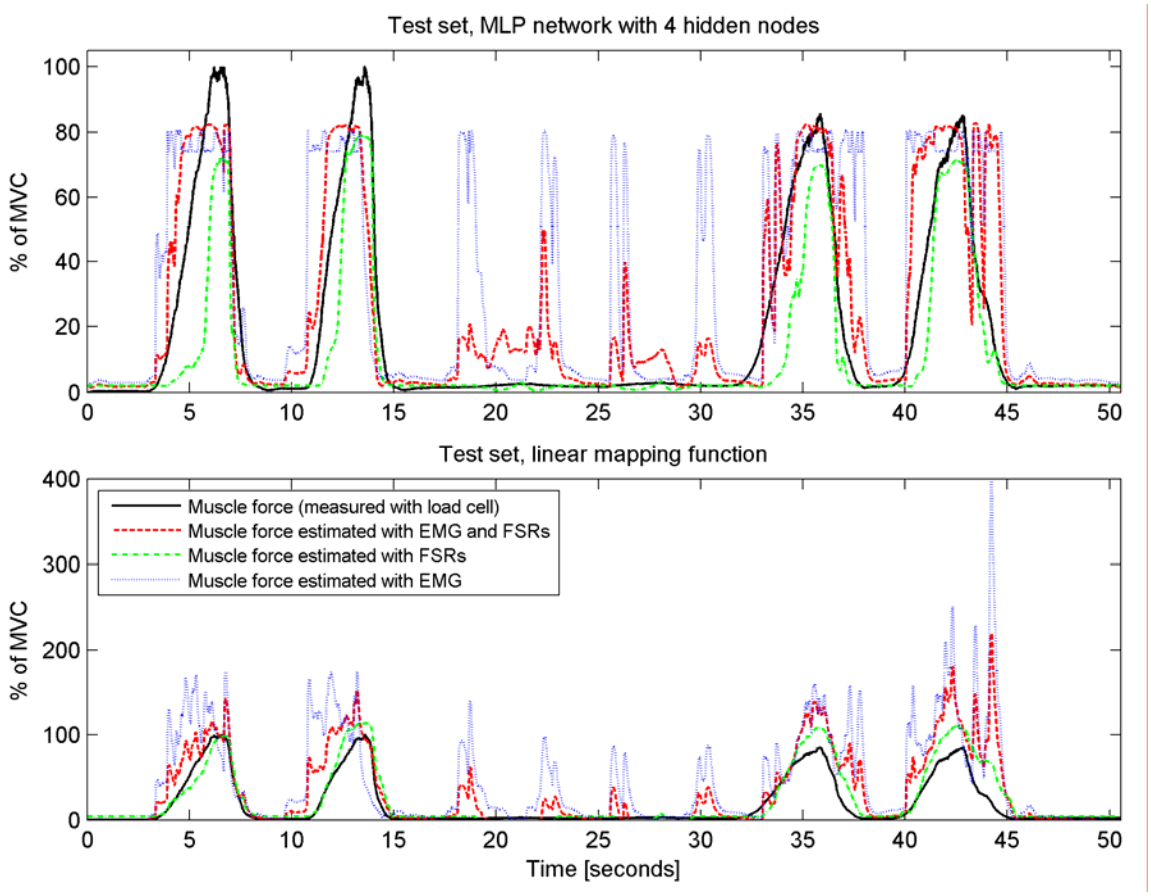

Fig. 3: Estimation results for three different test set inputs. Estimation using an MLP with 4 hidden nodes and a linear mapping function. Note different vertical scales; unit is percent of maximum voluntary contraction (MVC).

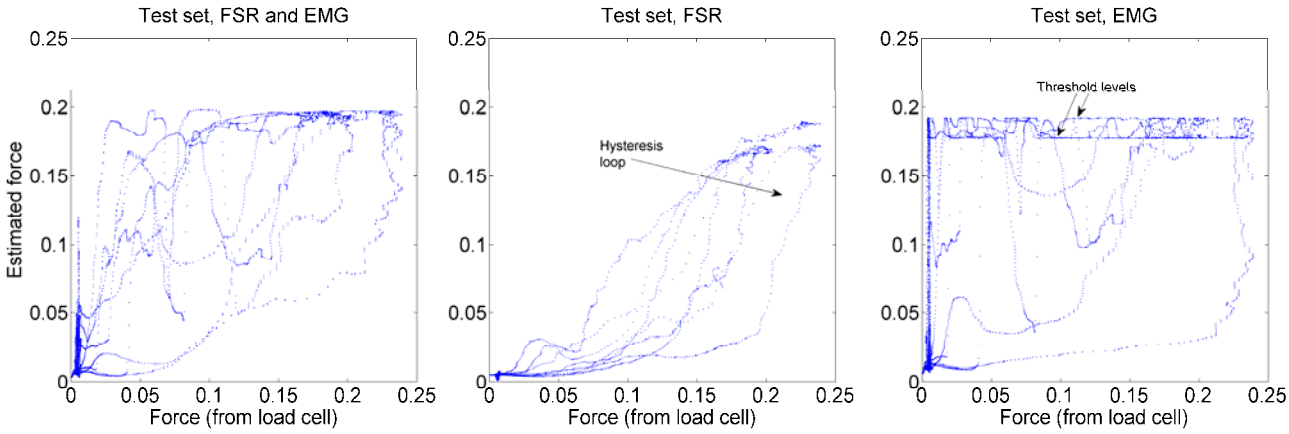

Fig. 4: Measured vs estimated force using an MLP with 4 hidden nodes. Same data set as in Fig. 3.

DISCUSSION

The results indicate that four hidden MLP nodes is a sufficient number to discriminate forces, as no improvement can be seen when increasing the MLP size beyond this point. The optimal MLP performed better than a linear estimator except when basing the estimate on FSR measurements alone, in which case the two techniques were equally successful.

The quadratic estimator was fitted to the training set without any validation. The results indicate that this has caused "overtraining" with respect to the training data, as evident from the estimator's inferior performance when subjected to the test set (cf. the caption of Fig. 5).

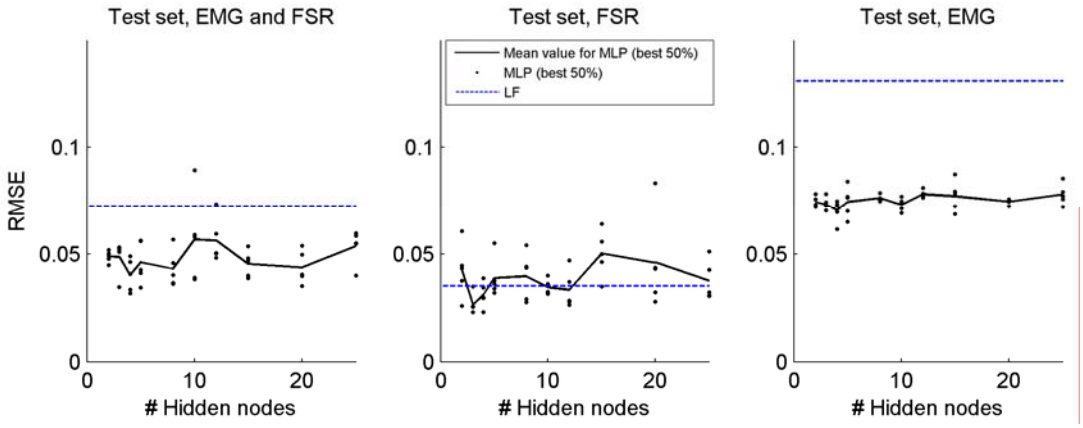

Fig. 5: RMS error rates for the same data set as in Fig. 3-Fig. 4. Corresponding values for the quadratic mapping function are 0.269 (EMG and FSR), 0.164 (FSR) and 0.152 (EMG).

It is noted that in Fig. 3 and Fig. 4, the FSR based estimates exhibit little or no artifact from external forces, which is at first a little surprising. In Fig. 2, however, it can be seen that the pure disturbance (i.e. the middle two "peaks") cause an equal response in all three FSRs, while when the muscle actually contracts, the FSRs yield different signal levels. Consequently, the estimator is able to distinguish these two signal sources.

In the upper graph of Fig. 2, the processed sEMG exhibits a transient response to the disturbance. This suggests that an optimal contraction force estimator should have a dynamic aspect rather than a purely static mapping property like the ones investigated in this study.

The results presented here are of a preliminary nature, and future study will assess the techniques using prosthetic sockets, real users and different myoprocessors. For example, the performance of a multi-FSR array inside a socket must be investigated, as the contact forces in that case may be different from those of the taped-on setup used in this study.

CONCLUSION

Measurements of contact forces exhibit promising properties for reducing force induced artifacts in conjunction with prosthesis control. The relative importance of sEMG and force measurements remain uncertain, and should be addressed in future work.

ACKNOWLEDGEMENTS

The authors would like to thank Professor Dennis Lovely of the Insititute of Biomedical Engineering, UNB, Fredericton, NB, Canada, for valuable input regarding the origins and nature of EMG artifacts.

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A VIRTUAL ENVIRONMENT ASSESSMENT OF A NOVEL PATTERN RECOGNITION BASED MYOELECTRIC CONTROL SCHEME

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ABSTRACT

This work compared a novel pattern recognition based myoelectric control system to a system based on conventional control and another state of the art pattern recognition system. The results showed that the proposed system provides a more usable system as assessed qualitatively and quantitatively through a modified virtual clothespin test. Furthermore, the proposed system was designed to have an intuitive clinician interface and should help facilitate the acceptance of pattern recognition based myoelectric control systems in the clinic.

INTRODUCTION

Pattern recognition based systems have demonstrated the ability to discriminate between many different degrees of freedom (DOFs) with a high degree of accuracy in controlled laboratory experiments; however, they have yet to receive widespread clinical acceptance. An intuitive configuration interface which provides feedback about the DOFs under control is a necessity for its acceptance. Furthermore, the control system must demonstrate good classification accuracy of intended motions and, more importantly, real-time controllability.

All pattern recognition based myoelectric control systems operate on the assumption that at a given electrode location, the set of features describing the myoelectric signal will be repeatable for a given state of muscle activation and different from one state of activation to another [1]. The problem is then reduced to representation of the MES signal, mapping the raw MES to feature space, followed by discrimination/classification of that space for the various motions. This work introduces a novel pattern recognition based myoelectric control scheme based on parallel binary classifiers. The system was assessed using a clothespin test implemented in a virtual environment and compared to a conventional control system based on mode switching and another state-of-the-art pattern recognition system.

BACKGROUND

Parallel binary classifiers have been used in previous pattern recognition problems [2]; however, not explicitly in the context of myoelectric control. The proposed control scheme, termed a multiple binary classifier (MBC) is based on uncorrelated linear discriminant analysis (ULDA) feature reduction [3] prior to one-versus-all classification [2].

Linear Discriminant Analysis (LDA) is a well known signal processing tool for classification and dimensionality reduction. Given a set of high dimensional data grouped by classes, LDA provides an optimal linear transformation to a lower dimensional manifold by simultaneously minimizing the within class distance and maximizing between class distance. ULDA further constrains the LDA algorithm such that the resulting transformations are uncorrelated, thus ensuring that the LDA algorithm converges even for the cases of non-singularity within class scatter matrices. A mathematical derivation and efficient algorithm for performing ULDA feature reduction was given by Ye et al [3]. An important property of the ULDA algorithm is that it will yield at most C-1 linearly independent features, where C is the number of classes in the data.

In one-versus-all classification, a separate classifier is made for each class, with classes being grouped as the target class and all other classes. This is best demonstrated through a simple conceptual example.

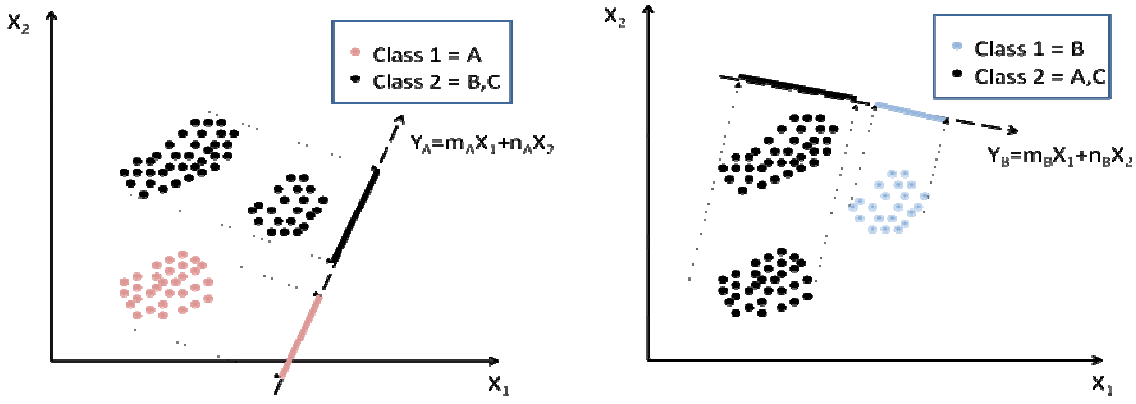

Figure 1: Conceptual Example of one-verses-all Feature Reduction. The classes (A,B, and C) are grouped into the desired class (Class 1) and all other classes (Class 2). The classifiers for classes A and B are shown. The classifier for Class C would be constructed in the same manner. The ULDA single discriminatory feature is also shown for each classifier.

When the number of classes is reduced to two using one-versus-all classification, ULDA feature reduction may be used to visualize the single discriminatory feature. If the classes are linearly separable, an adjustable threshold can be set to separate the data. The patterns are projected down the set of ULDA feature reduction matrices in parallel to determine the appropriate class as shown in Figure 2.

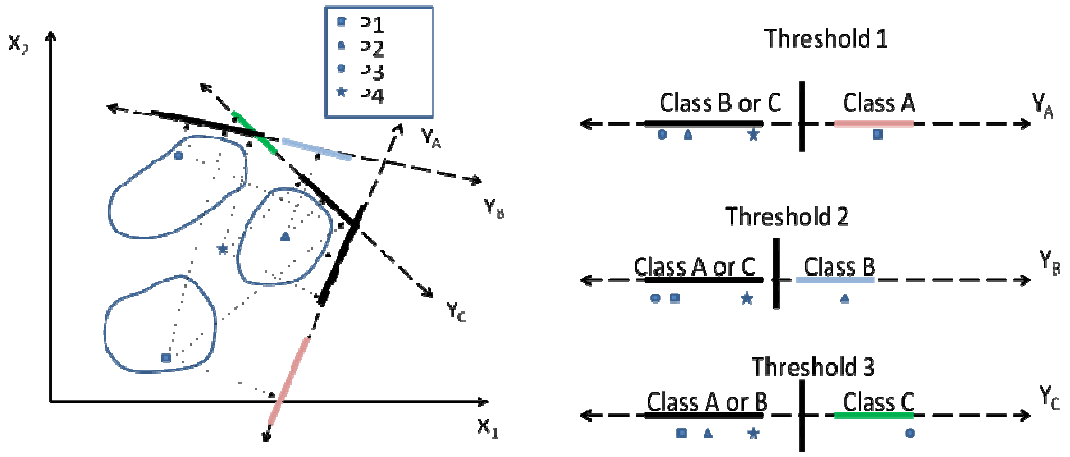

Figure 2: Parallel projections of patterns down the ULDA transformation matrices. Patterns located in the proper feature space regions are identified as the appropriate class.

If two or more of the binary classifiers produce outputs corresponding to their active class, there is a strong indication that the MBC has made an error and the safest course of action is to map the output to 'no motion'. The previous conceptual example is admittedly simple, but the technique should generalize to N-dimensional space provided the features are linearly separable.

METHODS

The proposed MBC classifier was assessed using a modified virtual clothespin test [4] and was compared to a conventional control (CON) system with mode switching and another state-of-the-art pattern recognition based myoelectric control system based on multi-class linear discriminant analysis (LDA) [5]. 12 normally limbed subjects completed the usability test with each of the control systems on separate days. The usability test required that the subjects have control of the wrist flexion/extension, wrist rotation, and hand open and close DOFs. For the conventional control system, electrodes were placed over the wrist flexor and wrist extensor muscle groups. A mode switch, operated using the subjects opposite arm, was used to toggle through the DOFs. For the pattern recognition systems, 8 electrodes were placed at equidistant locations around circumference of the forearm. Training and test data were collected in guided sessions using the UNB Acquisition and Control Environment (ACE) [6].

After the control system was trained, the user entered the virtual environment and was allowed to practice operating the limb. The clothespin test, a simulation of the real-world assessment technique, is a modified version of the one used by Lock [4]. Starting from a neutral position, the user is required to pick up a clothespin from a horizontal bar and move it to a vertical bar. When the hand is in position and closed over the clothespin, the pin turns red and is picked up. When the pin is positioned properly on the vertical bar, it turns green and the user can open the hand to release the pin. If the user opens the hand when the pin is not position, the pin drops and a new one appears on the horizontal bar. One clothespin test trial consisted of moving three clothespins from the horizontal bar to the vertical bar. The average time taken to move each pin and the total number of pin drops were recorded for each trial. The subjects were given one 'orientation' trial during which they were coached on where to position the hand to pick up and drop off the clothespin. After the orientation, subjects completed 8 clothespins trials with each control scheme. The metrics recorded were the average time taken to move clothespins during each trial and the number of clothespins dropped. The subjects were also observed and assessed qualitatively by the experimenter as they completed the test. Furthermore, offline classification accuracies were found for the pattern recognition systems to investigate a relationship between classification accuracy and usability.

RESULTS AND DISCUSSION

Table 1 displays the average pin placement times, the standard deviation of the pin placement times between subjects, and the total number of pins dropped and classification results for all subjects for each of the control schemes.

	Control Scheme		
	CON	**LDA**	**MBC**
Placement Time (s)	28.6	27.2	21.4
STDev Time (s)	8.7	12.1	7.4
Total Drops	8	29	1
Classification Error (%)	N/A	2.6	4.9
STDev Error (%)	N/A	1.7	2.4

Table 1: Results of the Virtual Clothespin Usability Study

It should be noted that results presented are for trials 5-8 for each subject because the subjects exhibited a learning effect during the first 4 trials, but very little learning thereafter. A statistical ANOVA showed that the MBC had significantly lower pin placement times (**p<0.001**) but a significantly higher classification error (**p<0.001**). A statistical analysis was not completed on pin drops, but it is apparent that the MBC classifier yielded fewer pin drops than the CON and LDA systems.

A scatter plot of the classification error and pin placement times is shown in Figure 3.

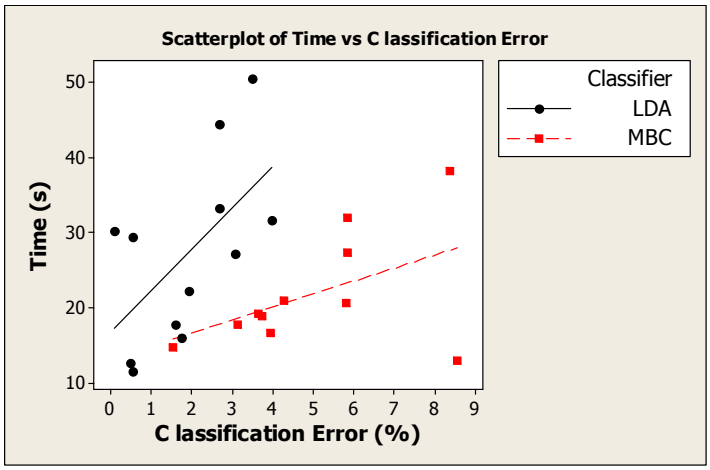

Figure 3: Scatter plot of average placement time vs classification error for the pattern recognition systems.

A positive correlation between classification error and pin placement time was observed for both control schemes. The main observation was that the classification error metric for each classifier is computed in the same manner; however, the slope of the regressions lines differs. This implies that classification error does impact usability but that there are additional factors as well. The number of data points in the scatter plot is limited and addition data should be collected before stronger conclusions may be drawn.

The following paragraph summarizes the qualitative observations made during the experiment. Firstly, subjects disliked the mode switched conventional control because they found it difficult to keep track of which mode was next in the sequence. Subjects could easily operate the DOF once they were certain which mode they were in. Another criticism of the conventional control scheme was that contractions were not physiologically appropriate and some had difficulty remembering that wrist flexion could operate hand close or wrist pronation depending on the active mode. Subjects generally preferred the MBC and LDA control schemes over the conventional based system. Most subjects found it more intuitive to use because the control was physiologically appropriate and a mode switch was not required. The MBC control scheme had a higher classification error, but in all cases the majority of errors were 'no-motion' errors meaning the prosthesis did not operate. The majority of LDA classification errors were errors to other 'active' classes. These inadvertent activations were very problematic for most subjects and led to slower pin placement times because the subject had to correct the error before continuing with the test. Inadvertent activations were also to blame for the majority of pin drops noted with the LDA classifier; the hand would often involuntarily open when the subject was trying to extend the wrist. All three control systems were criticized for not providing

simultaneous control over multiple DOFs. The virtual environment was also criticized for a lack of depth perception and slow rendering time.

CONCLUSIONS

The MBC control scheme is a novel pattern recognition based myoelectric control system which was designed to have a clinically intuitive interface. After training data is collected, the clinician can set a series of thresholds for each DOF in a similar manner as is done with current conventional control systems. The MBC control system yielded faster clothespin placement times when compared to a conventional control based system and another state-of-the art pattern recognition system. Furthermore, the majority of the subjects stated a preference for the MBC control system during usability testing.

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SMALL, LIGHT-WEIGHT, BUILT-IN BATTERIES OFFER MORE ENERGY AND IMPROVED COSMESIS

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Liberating Technologies, Inc

FINDING SMALL LIGHT-WEIGHT CELLS

One might think that with over a hundred Li-Polymer cells on the market, one could find just the right cell for every application, but this is far from the case. Most cells are made in Asia by companies that only sell in large quantities. Dimensions are driven by the size and shape that is appropriate for 3.6V hand held devices. For these devices thin and flat are paramount, while the linear dimensions are driven by the size of the device and are usually too wide for a prosthesis. Figure 1 shows four cells that have proven usable in prosthetics. Left to right, the first cell is a mere 200 mAh, but is really small. The second, 250 mAh, is the smallest rectangle that will mount crosswise at the end of a child prosthesis. Unfortunately, it has been discontinued by the manufacturer. The third cell, 450 mAh, is slim in two dimensions so it fits well with one cell each on side of a wrist. The 750 mAh cell on the right is good for a full day's operation of an adult transradial prosthesis.

Table 1. The Ideal Built-in Battery has:

Light weight	Convenient shape
High energy density	User safety
Low internal resistance	Lasts a whole day
Ease of recharge	Ease of installation
Small size	Separable components

Cells must be selected for their electrical characteristics as well as for size. For instance, many of lithium ion cells have a high internal resistance limiting the current that can be drawn. Recent changes in the chemistry of some of the lithium polymer cells have opened up new high current applications. To move an elbow or shoulder with weight in the gripping device requires substantial current even when extra cells are added to increase voltage. With battery operation every component must be as efficient as possible and this includes the batteries.

Figure 1. Typical Li-Poly cells used in LTI batteries before adding protective circuits. Capacities are 200, 250, 450, and 750 mAh.

Figure 2. A pair of 750 mAh cells (30 mm wide, 44 g) compared with larger pairs 1350 mAh (33 mm, 66 g) and 2100 mAh (36mm, 88 g) and a 2000 mAh cylindrical pack (Two 18 mm cells, 98g).

SAFETY WITH LITHIUM BATTERIES

Connecting the Charger Safely

Myoelectric prostheses usually ground the electrodes to the negative side of the battery. This can be a problem if it is possible to connect the battery to a source of high voltage. This might occur in a fault condition if charging while wearing the prosthesis. As long as the prosthesis is turned on, it must be impossible to touch any metal surface that is positive. (An infant once took a nap while skin was touching metal at a positive voltage and received a minor burn.) Isolating

both the cells and the external connector is the function of the recharge module. Older modules used a double pole double throw switch for this. To save space, newer modules use a switch in the recharge connector to shift the negative side of the battery from the user electronics to the outside of the jack when a plug is inserted. The single pole single throw on-off switch does the same for the positive side, as the switch is moved from on to recharge.

Electrical Safety for the Cells

During assembly, two safety features are added to all LTI lithium cell packs. In figure 3, the safety circuit on the left monitors the voltage across both cells of a 750 mAh cell pack. Its most important function is to prevent either cell from being over charged. It also prevents the voltage from dropping too low by disconnecting the cells if the voltage drops below a specified value or if a short occurs. The right cell has a Poly-Fuse™ installed to limit current in a short circuit across just this cell as might occur if the connecting cable were shorted. Also, since a Poly-Fuse™ is a device that changes state with temperature, it may act as a thermal current limiter. The safety devices are mounted close to the center line so that the added thickness can be ignored when the cell is placed along side the concave surface inside a prosthesis.

Figure 3. Safety features of a typical battery. The cell on the left mounts the safety circuit that monitors each cell separately. A Polyfuse on the right limits short circuit current and prevents over heating.

Making Cells Fit

The human forearm is oval just proximal to the wrist and then becomes more or less round toward the elbow. However, it is traditional to make the interface between the terminal device and the forearm round. Thus, there is room for a thin cell on each side of an adult wrist without increasing the overall size. Figures 4 and 5 show that there is plenty of room for the cells supplied in the LTI 450 and 750 mAh batteries. For comparison figure 6 shows how the larger cells that can handle the requirements of the i-Limb™ hand may interfere with cosmesis on a typical installation. The cells almost fit at the wrist, but they are 100 mm long and will cause lumps at the proximal ends.

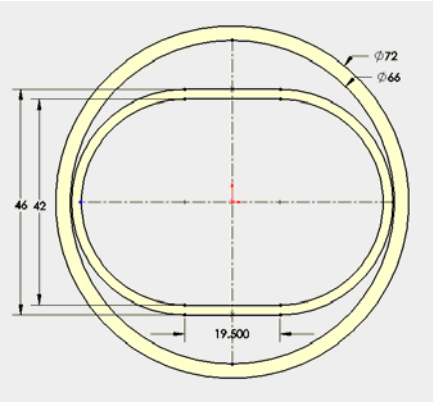

Figure 4. Typical dimensions for a male forearm 80 mm above the wrist.

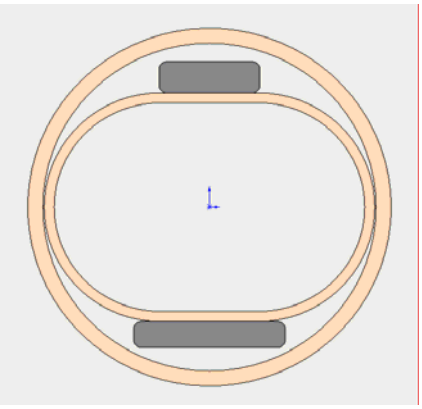

Figure 5. Cross sections: On top, a 450 mAh cell; below a 750 mAh cell.

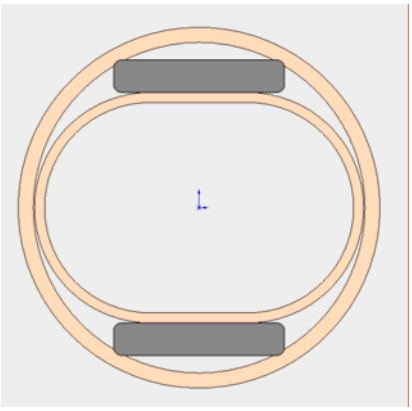

Figure 6. Two 2100 mAh cells for an i-Limb weigh 88 g and need more room.

Safety When Placing the Cells

Lithium polymer cells are covered by a thin membrane which provides little protection from sharp objects. Furthermore the cells lack resistance to crushing and bending. The shrink wrap added during assembly only helps a little. The technician must use dummy cells when making the definitive outer socket to provide space for cells, cables, and for the recharge modules. If the Recharge Module is located remotely, there will be wire-to-wire connectors that will need a cavity. Careful planning is required so that the final prosthesis can be readily assembled and serviced. One of the most hazardous times for the battery components is during installation when there is a great temptation to push wires and connectors into too small a space.

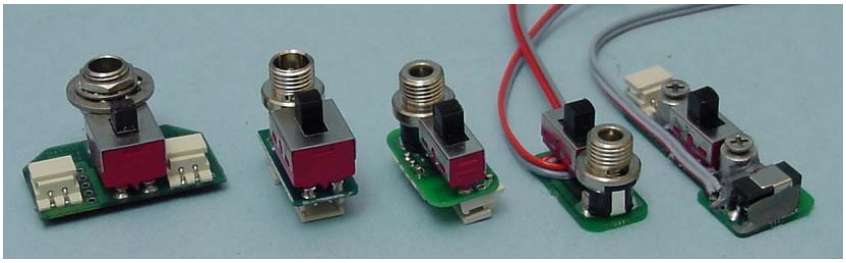

Figure 7. Recharge connectors and switches have gotten smaller. From the top of the switch the units measure 13, 16.7, 14.7, 11, and 8.3 mm in height. A thin recharge assembly can be placed near the elbow on the forearm

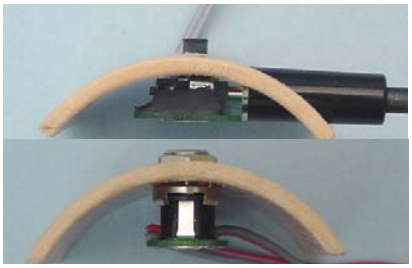

Figure 8. The 8.3 mm module top vs. the 11 mm below, both on samples 57 mm OD by 2.6 mm thick.

Robust Recharge/On-Off Modules

Figure 7 shows two older LTI Recharge Modules and three newer ones. All are designed to withstand the rigors of installation in a typical prosthesis while maintaining electrical safety. Sometimes local shops invent their own modules to save space, but the compromises are not acceptable in a device offered for sale. Small switches will not repeatedly switch high currents nor will miniature connectors hold up when one handed users insert and retract recharge plugs.

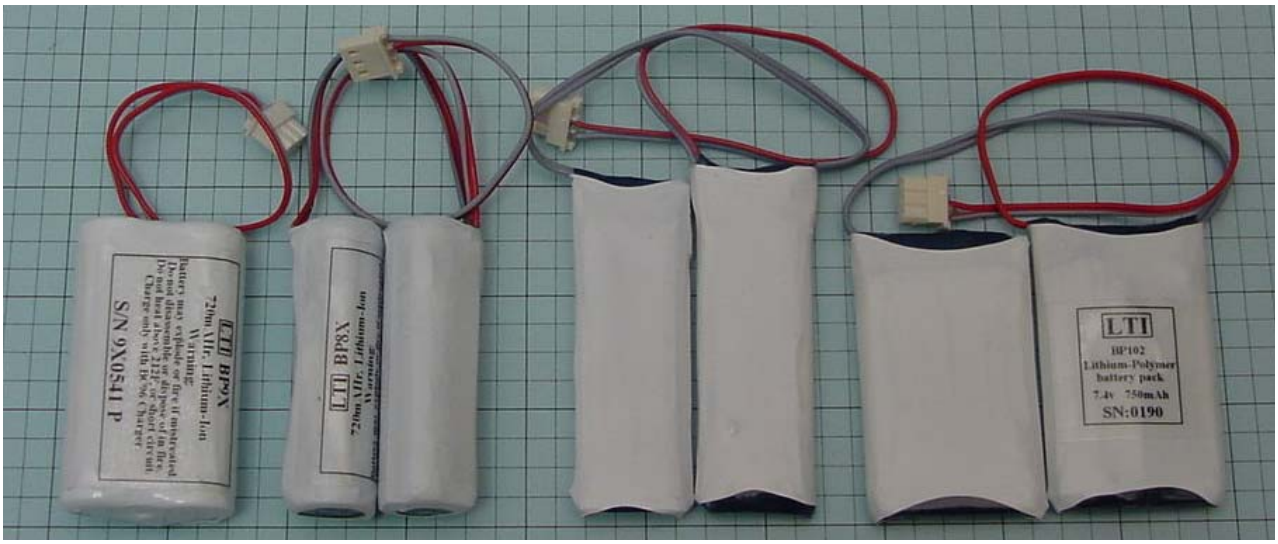

Figure 9. LTI cell packs come in several styles. The pack made from two AA Li-Ion cells is robust and remains popular. Separate AA cells are often stuffed into fingers of large hands. The newer flat, thin Li-Poly cells can be placed on opposite sides of the wrist or one above the other lengthwise.

ELECTRIC UNLOCK FOR LTI LOCKING SHOULDER JOINT

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THE MANUAL LOCKING SHOULDER JOINT

At the 2002 MEC, LTI introduced the improved Collier Shoulder Joint which was still being shown at the 2005 MEC. In 2006, the Collier design was replaced by a completely new joint with greatly improved durability. The original joint had an option of adding on an electric unlocking mechanism that was slow and noisy. A year later in the summer of 2007 an electric unlock was made available for the new joint. This paper will discuss the design features of this new shoulder joint and the kit that can convert it to a joint that is locked and unlocked by an electric actuator.

Figure 1. The LTI SJ90 Manual Lock Shoulder Joint.

Table 1. Design Goals for the 2006 LTI Locking Shoulder Joint

The flexion-extension axis will have two states – locked and free to swing.
The adduction-abduction axis will have adjustable friction.
Both axes will sustain at least a 50 ft-lb (68 Nm) load.
A center hole will accommodate wires and connectors.
The lock will engage every 10 degrees.
The lock will alternate between states like a retractable ball point pen.
A spring will provide delayed locking if the joint is not initially aligned.
Both size and weight will be minimized while meeting all other goals.
The mechanism should serve a heavy user for five or more years.
The lock must operate manually or with a Bowden cable and remote lever.
Corrosion resistant materials will be used throughout.
Out of fairness to those who had the Collier joint and might want to upgrade:
- The bolt circle and screw spacing for mounting will be unchanged.
- The mounting ring will be the same size.
- The Collier humeral plate will fit the new joint.

In less than two years several hundred SJ90 Shoulder Joints have been delivered. The first joints were sent to users who had had difficulty maintaining the older joint. These tests uncovered several weaknesses which were immediately corrected. Two other structural problems later occurred at the 2% level. They have been addressed in the 2008 run. Important features of the mechanism are in figures 2-3. While the joint is relatively thin and only 58 mm in diameter, it will withstand heavy loads. The ball races are far apart, and the locking posts are on a large diameter circle. Furthermore six plungers share the load when the lock is engaged.

Figure 2. Unlock cam ring drives six plungers. The large pin is driven by the puller.

Figure 3. Six plungers move outward to engage slots between 36 posts in rotor. The large diameter ball races withstand the full 50 ft-lb (68Nm) design torque.

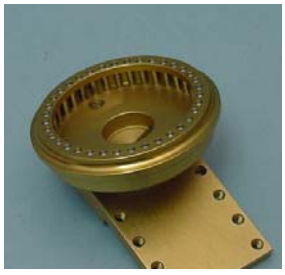

Figure 4. The 36 posts in the rotor are hardened stainless steel dowel pins.

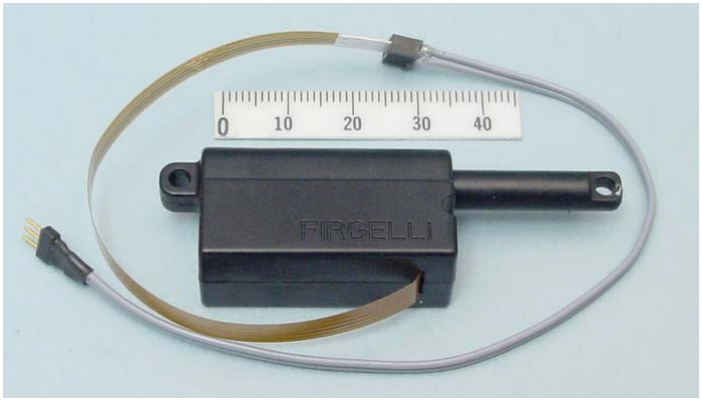

Figure 1. The Linear Actuator
Stroke ─ 20 mm
Force at peak efficiency ─ 15 N at 7 mm/s
Max Power Point ─ 18 N at 6 mm/s
Max speed (no load) ─ 12 mm/s
Current draw at peak efficiency ─ 250 mA
Input voltage ─ 5 V DC
Feedback mode ─ 2 kΩ linear pot.
Feedback potentiometer linearity ─ 1%
Mass ─ 19 g
Operating temperature ─ –10 to +50°C

DESIGN GOALS FOR THE ELECTRIC UNLOCK KIT

It was surprisingly difficult to specify the design of the electric unlock because many requirements could only be added by dropping others. For instance, a cable puller to activate the reciprocator on the manual unlock joint would not fit in a convenient space. The final design is integral to the joint.

Table 2. Design List for the Electric Unlock

It must go with other manufacturer's batteries and controls.
It must be easy to change a manual joint into an electric unlock joint.
No extra control circuitry should be required.
All actuators, motors or gears should be off-the-shelf-items.
A spring between the actuator and lock for delayed lock or unlock.
The actuator must be quiet and able to act in a second or less.
Must fit below the joint in the same area as the manual-joint actuator.

New Actuator Becomes Available

A breakthrough in the design came with the availability of the miniature hobby linear actuator shown in figure 1. It comes in both high-speed and high-force versions, and the latter proved to work best. Mechanical tests showed that the new actuator could both lock and unlock the joint mechanism by pushing or pulling a spring similar to the one used in the manual lock joint. The spring is attached directly to the puller that moves the lock. Figure 5 shows all the parts. The actuator only needs to move 9.6 mm between the positions where the drive motor stalls.

The actuator has a simple plug to attach to any LTI cable set. Cables are offered with connections for most of the available prosthetic switches and standard 7.2V batteries. A 5V regulator is placed between the battery and switch so that the actuator is never overdriven.

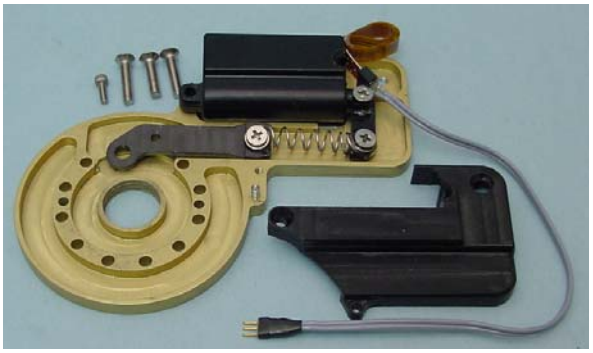

Figure 5. The actuator moves only 11 mm to move the puller left or right. When the puller is stuck the spring takes up the motion and stores it for later release.

Figure 6. The base plate supplied with the manual joint on the left is held in place by only two screws making it easy to substitute the electric unlock base on the right.

CONDUCTIVE INSERTS TO ACQUIRE MYOELECTRIC SIGNALS THROUGH SILICONE LINERS

William J. Hanson
Liberating Technologies, Inc.

INTRODUCTION

Prosthetic socket liners provide both suspension and stability, and they do so best when they cover the remaining limb without breaks in the liner surface. This full coverage conflicts with the need to acquire myoelectric signals directly from the skin. Early solutions to this problem include windows in the liner and metal electrodes piercing and thereby weakening the liner. Windows compromise the integrity of the liner and often lead to discomfort at the edge of the window. Metal electrodes require attachment of external wires after donning. Another alternative is to mount cased electrodes in molded silicone receptacles. This may result in a proper seal, but it requires repeated removal of the electrode from the liner.

The problems identified above can be addressed by passing the myoelectric signals directly though the liner with flexible Conductive Inserts. This approach allows metal electrodes in the outer socket to acquire the myoelectric signals as if they were in direct contact with the skin with no wires attached to the liner. LTI has developed Conductive Inserts that not only pass myoelectric signals through the liner but also allow for misalignment when the liner is inserted into the socket. These Inserts can be installed by prosthetic technicians in a commercial liner, or they can be built into a custom liner. With Inserts installed, a liner is a separate part that can be easily removed for cleaning.

ELEMENTS OF THE INSERT SYSTEM

The system is quite simple. For two electrode sites, five or six of the Inserts shown in figure 1 are installed in a production or custom silicone liner with the wide “overlap” on the outside. The center part of each Insert is 12 mm in diameter and either 2mm or 3mm thick to glue into a hole punched in the liner. The larger diameter overlap is about ½ mm thick and provides a greater target area for the Metal Electrode. The overlap also provides increased strength because adhesive bonds are strongest in shear.

The resulting liner with Inserts is shown in figure 2. In the meantime, a clear check socket is used to verify the fit of the liner. Once the patient can don the liner and place it into the check socket in a consistent manner, the centers of the Inserts are marked on the check socket for installation of the Metal Electrodes shown in figure 3 (three sizes available). The result is shown in figure 4 where the three cables from an LTI Remote Electrode are connected to metal electrodes. Note that the electrode may be slightly offset from the center of the Insert, but still make contact. The final element is the Remote Electrode itself, shown in figure 5.

Figure 1. Conductive Insert

Figure 2. Liner with Inserts installed

Figure 3. LTI Metal Electrode sizes

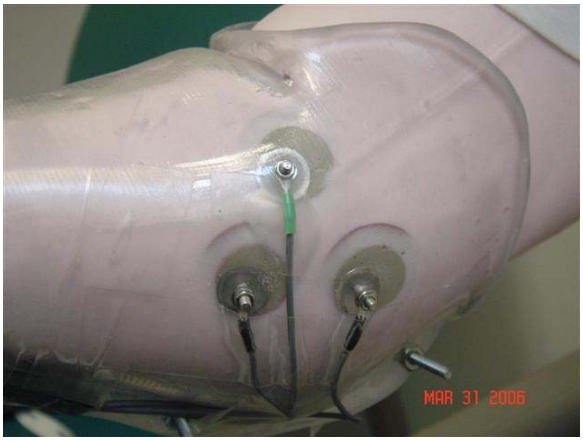

Figure 4. Remote Electrode Cables attached to the Metal Electrodes on the check socket.

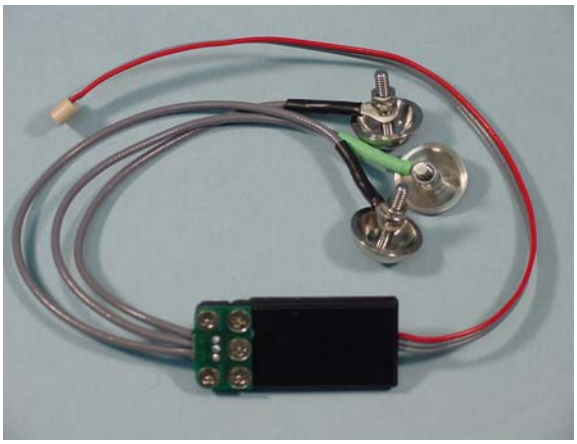

Figure 5. Complete LTI Remote Electrode System with a connector for a Bock or Steeper wrist.

OPTIONS FOR SUPPLYING SYSTEMS

Option 1: The Inserts are installed by the supplier

The prosthetist determines the residual limb measurements and a correctly sized silicone liner is sent to the facility. The patient dons the liner and the practitioner marks the myoelectrode sites. The liner is then sent back to the central fabrication facility where the Inserts are installed. The patient's liner with the Inserts is returned to the practitioner along with the appropriate Remote Electrodes. At this point, the practitioner builds a check socket and mounts the Remote Electrodes to align with the Inserts in the liner. Final testing is done to assure good myoelectric signals, and these electrode locations are then transferred to the definitive socket. Most early fittings have been done using this option.

Option 2: A kit for local installation of the Inserts

As an alternative to central fabrication of the Inserts into a production liner, LTI plans to provide these Conductive Inserts as a kit for the local technician to install in a production liner or for installation in a custom silicone liner, thus reducing the turn-around time. The kit will consist of 3 Conductive Inserts, silicone adhesive and a mounting template. At present, we are experimenting with various manufacturers' upper-extremity silicone liners to insure that the selected adhesive works well. On a trial basis, the proposed kit has been used by two overseas practitioners where shipping back for fabrication would have been prohibitive. The preliminary results with custom liners looks promising. LTI is awaiting the final results before offering this option.

Future options

During field trials with the Conductive Inserts, it became apparent that some applications may need smaller Inserts. In the future, experiments will be done with Inserts measuring 5 or 6 mm in diameter. This reduced diameter will permit a smaller overlap, but will give a more secure bond. However, the diameter of the overlap must not get too small or it will become difficult for users to align the Inserts with the metal electrodes. The small Inserts will be useful for the close muscle sites required for partial hands, Targeted Muscle Reinnervation patients or for sampling multiple sites for use with pattern recognition/classification schemes.

QUALITY FOR LIFE

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Michelangelo 03 - A versatile hand prosthesis, featuring superb controllability and sophisticated bio mimicry

(Abstract)

Rapid progress has been made recently in all components of upper limb myoelectric prosthetics, and revolutionary new concepts in controlling these devices are part of ongoing R&D activities. Nevertheless there is a need for new prosthetic hands capable of providing a whole new set of functions.

These new capabilities, combined with well proven features like robustness and reliability, together create new standards in upper limb prosthetics.

The Michelangelo Hand Prosthesis raises the controllability in the areas of both speed and precision. Having motion speed and grip force which are virtually the same as the natural human equivalent will allow much easier control for the user. An emphasis on smoothness in controlling the motion of the fingers and the thumb will guarantee sensitivity and precision.

The biggest modifications are related to a higher number of DOF within the Michelangelo hand prosthesis. Multi-axial articulation of different hand and finger segments allows a higher number of gripping patterns, and the resulting high adaptability improves the overall performance of ADL's. In order for the prosthesis to emulate the natural human hand special effort and focus was placed on bio mimetic design and shape profiling.

TECHNOLOGY FOR EVALUATION, FITTING AND TRAINING DURING THE PROCESS OF UPPER LIMB PROSTHESES MANUFACTURING

Axel Eichinger[1], Andreas Rogge[1]
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EXECUTIVE SUMMARY

The paper outlines that a huge diversity of components and different control modes are necessary to satisfy patients' needs. Therefore, satisfaction is the critical factor whether a patient will wear his prosthesis or not. Consequently, technological effort is directed towards research and development departments to supply a product that can handle the amount of different configuration possibilities.

PAULA will assist certified prosthetists with all steps involved in manufacturing arm prostheses such as patient management and documentation, muscle signals evaluation, component selection and ordering service. In addition PAULA will provide beneficial features for users that will enhance both the quality of orthopedic fittings and the individual patient care. The complexity and opportunities of the product is a milestone in the field of arm prosthetics.

IDEA

The demand for better adaptability to individual patient needs has increased the variety of available components as well as the complexity of modern upper limb prostheses considerably. Several manufacturers offer many components and different programs for Myoelectric, BodyPowered and Hybrid prostheses as well as cosmetic fittings. This multitude is very complicated and needs a lot of knowledge and experience.

Building prostheses requires the consideration of many patient related influencing factors (size selection, myosignal quality, etc.) as well as knowledge of technical details (e.g. component sizes, compatibility of various components, and available control options). Simultaneously, the usability and acceptance of prostheses not only depend on selection of proper components and their optimal adjustment but also the correct fit of the socket is crucial.

With regard to these circumstances, the development of PAULA was initiated. It should handle the enlarged complexity of components for the upper extremity and provide an E-learning platform for the whole MYOBOCK® system. Additionally, the new software tool should guide certified prosthetists through the whole process and helps them to choose the best components and improve the outcome of the fitting. Occupational therapy can also be supported by the help of the new software.
There is no doubt: This software adds value for CPO, OT and patient.

STEPS OF DEVELOPMENT

In general, the development of Paula was realized in three fundamental steps:

1. In 2003 the computer assisted fabrication service (CAFS) team, an Otto Bock department for customer oriented software programming, was assigned to develop a new software tool depending on the requirements of the target groups. In other words, this software system should contain a patient management tool and a component selector for Otto Bock upper extremity products. Moreover, the software should assist prosthetists with the socket design for transradial amputees. The CAFS team essentially used the knowledge and experience gathered from the computer aided design for sockets for lower extremities. This know how was altered and adjusted to the circumstances of transradial amputees.

2. In 2004 the Otto Bock research and development department in Vienna, Austria, started to expand the existing MyoSoft®. This software program was used for testing and evaluation of patients' muscle signals and the subsequent myo-simulation. The idea of this project was the establishment of a new patient management and the introduction of new features of the MyoSoft® system in a new layout.

3. The final step was to join these parallel proceeding projects together to one common product. As a result, it was agreed that the features MyoSoft, socket design, patient management, component selector, and ordering service should not run as a stand alone product but rather as an integrated software solution with a common user interface and platform that permit easy handling and efficient workflows.

DESCRIPTION OF THE PRODUCT

This abbreviation stands for **P**rosthetist's **A**ssistant for **U**pper **L**imb **A**rchitecture. The product is a complex software program that supports certified prosthetists during planning, designing and manufacturing prostheses in the field of upper extremities. It can be used for both Myoelectric- and BodyPowered- prostheses for all levels of amputation as well as for passive arm prostheses. PAULA consists of five different modules integrated in one common platform. In detail, these applications are described as following:

1. Patient management

Manufacturing prostheses requires much information related to the patient such as personal details, amputation level and side, or gender. The patient management of PAULA uses these indications for documentation purposes. It also includes the administration of pictures of the patient and other patient related information that can be saved. Different jobs can be created for the same patient. This tool is beneficial for the justification of a new prosthesis because changes at the stump can be archived and therefore demonstrated.

2. Transradial socket design

Socket design is essential for both the fitting of prosthesis and the satisfaction of the patient. With the help of PAULA the fabrication of sockets for patients with transradial amputation levels can be realized without taking a plaster cast. The certified prosthetist takes pictures from the residual limb of the patient, digitalizes the socket and uses the software to

modify the socket in the personal computer. As a result, he gets a 3D socket design that can be sent online to an Otto Bock service fabrication for manufacturing a test socket.

3. Myotest, Training and Simulation

Myoelectric prostheses are controlled by muscle signals that can be measured on the skin of the patient in case of muscle contraction. Therefore, electrodes are required to measure the muscle tension. The muscle potentials will be transmitted to the computer screen where the prosthetist can choose from different programs and component depending on patients' indication. PAULA enables the certified prosthetist and the patient to see the movement of the prosthesis in real-time on the screen. The certified prosthetist can evaluate the muscle potentials and test the best components' configuration. The simulation helps the patient to understand the different functions and to train his muscles.

4. Component Selection

Depending on the indication and the muscle signals of the patient the system make proposals for suitable components and control modes where the prosthetist can choose from. This component selection guarantees component compatibility for the entire prosthesis. However, the expert mode allows certified prosthetists to determine another component configuration for the patient.

5. Ordering

The final step in manufacturing prostheses is the purchase of required components. With regard to the four prior steps, PAULA creates an automatic order form that can be sent via email to the Otto Bock service fabrication in order to receive a test socket with all functional components integrated in an alignment tool for a complete upper limb test prosthesis. Instead of receiving an entire test prosthesis, the certified prosthetist can also order the components and assembles the prosthesis by himself.

CONCLUSION

To sum up these points, PAULA is a composite software program that offers a guide line for experienced certified prosthetists to assemble prostheses for upper extremities with complex control modes in the right configuration. It provides an explicit component selection and an easy ordering procedure. Furthermore, PAULA supplies mostly automatic documentation features and patient management tools.

Finally, it simplifies and speeds up all tasks from muscle testing and evaluation, socket design, muscle training and simulation, till the appropriate configuration of the prosthesis. As a conclusion, PAULA assists, supports and submits proposals and recommendations. With regard to these benefits, the quality of arm prostheses supply will enlarge. Additionally, PAULA will improve individual patient care. And for this reason the customer satisfaction will increase.

PROSTHESES CONTROL BASED ON TMR
a case study

Hans Dietl, PhD
Otto Bock Healthcare Products GmbH

Introduction

The surgical treatment by targeted muscle reinnervation has been applied to more than 25 patients after traumatic loss of upper extremities in North America.

In November 2006 this procedure was applied for the first time to a patient in Europe. The 18- year-old male patient suffered from bilateral limb loss after a high voltage accident, which happened in October 2006. The patient ended up with a medium long transhumeral residual limb on the right side and a shoulder disarticulation on the left side.

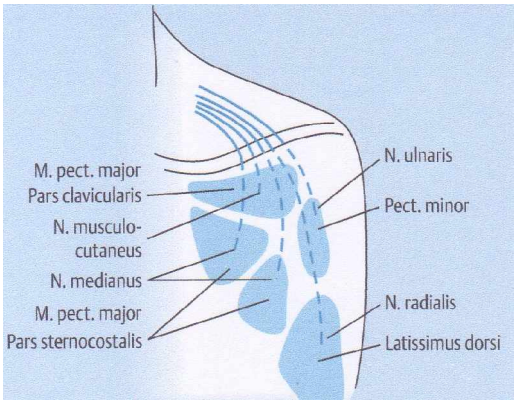
He did undergo a primary rehabilitation at the rehabilitation center "Weisser Hof" and left rehabilitation after three months. The right side was fitted with a DynamicArm®, a wrist rotator and a SensorHand Speed. The system was controlled by the EMG signals of the biceps and the triceps. The switching between levels was achieved by co-contraction. The left side was fitted by a cosmetically arm with minimal functionality since trial fittings with functional arms failed.

Assessment Phase

The initial rehabilitation outcome was not in accordance with the specific needs of the patient: he did not reach a sufficient level of independence for everyday activities and it was not possible for him to return to work. For an improved functional outcome the surgical procedure of targeted muscle reinnervation was considered. For an investigation about the local situation of the arm - plexus a screening by ultrahigh resolution ultrasound was done. This showed intact branches of the fascicular structure. Electrical diagnostic methods indicated intact proximal branches of the nerves. An interdisciplinary team, consisting of neurosurgeon, a Doctor for physical medicine, an occupational therapist, a CPO and some expert engineers for artificial limbs was formed. This team developed a rehabilitation plan. For the surgery a nerve transfer matrix was developed, which should secure proper functionality for the upcoming prosthetic fitting [Figure1].

Source (nerve)	**Target (neuromuscular unit)**	**Target phantom function**
N. musculo-cutaneous	N. pectoralis clav.	Elbow flexion
N. medianus	N. pectoralis stern.	Fingerflexion Pronation
N. medianus	N. Pectoralis abd.	Wrist flexion
N. ulnaris	N. Pectoralis minor	Intrinsic Function
N. radialis	N. thoracodorsalis	Fingerextension

Table 1: nerve transfer matrix



Surgical intervention

The goal of the intervention was to unhinge the greater branches of the arm plexus and to connect them to existing neuromuscular units. This rerouting was done according to figure1.

The nerve transfers were done by end to end nerve-cooptation. For better signal separation the musculus pectoralis minor and the M pectoralis major pars clavicularis were moved.

In addition the trunctus medius C7 was connected to the nervus supraclavicularis, which was cut at the punctum nervosum. By that means a sensible skin area should be generated, which can represent the sensory areas of the hand. The whole region was defatted for better signal quality.

Postoperative phase

Six weeks after the intervention the patient did undergo a training plan. He was supervised by an occupational therapist. The primary goal was to enhance the overall fitness and endurance and to correct the body posture and symmetry. As soon as measurable EMGs were detectable a specific training program was started. The patient visited the clinics every six weeks for follow-up. After three months voluntary muscle contractions at the M. pectoralis minor et major could be identified. Sensibility at the Trigonum colli laterale recovered. There the patient could clearly differentiate different regions of his hand. (Figure 2).

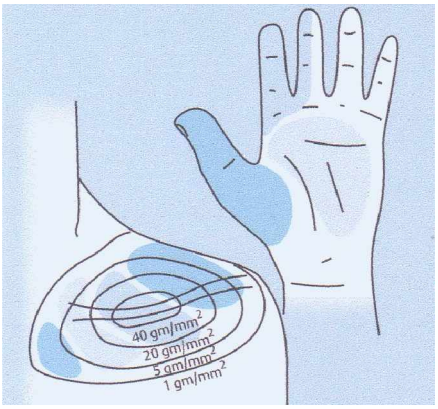


Figure 2: isobaric lines of sensation and hand regions

After six months there were sufficient signals at the segments of the M pectoralis, which allowed starting the fitting process (Table 2)

Source (nerve)	**Target (neuromuscular unit)**	**Result**	**Target phantom function**
N. musculo-cutaneous	N. pectoralis clav.	excellent	Elbow Flexion
N. medianus	N. pectoralis med.	excellent	Close fist
N. medianus	N. Pectoralis inf.	good	Wrist Flexion
N. ulnaris	N. Pectoralis minor	??	Intrinsic Function
N. radialis	N. thoracodorsalis	??	Finger Extension

Table 2 nerve transfer matrix postoperative after six months

After 17 months the nerve growth process seemed to be completed. All targets provide excellent signals. On the target units a further differentiation of phantom arm movements was possible (Table 3)

Source (nerve)	**Target (neuromuscular unit)**	**Result**	**Target phantom function**
N. musculo-cutaneous	N. pectoralis clav.	excellent	Ellbow Flexion
N. medianus	N. pectoralis med.	excellent	Close fist
N. medianus	N. Pectoralis inf.	excellent	Wrist Flexion, Pronation
N. ulnaris	N. Pectoralis minor	excellent	Intrinsic Function
N. radialis	N. thoracodorsalis	excellent	Finger Extension, Supination

Table 3 nerve transfer matrix postoperative after 17 months

Prosthetic fitting

The patient was fitted with two different arm systems, the “Take Home Arm” and the “6 Degrees of Freedom Arm”.

The “Take Home Arm”

This system has a passive shoulder joint with free swing and a preflected position, a modified DynamicArm®, a wrist rotation unit and a Sensorhand Speed. For direct control of the 3 Degrees Of Motion a sixth EMG signal was needed. Therefore the musculus deltoideus was integrated into the control scheme. Standard EMG electrodes were used, however because of the skin movement a modified suspension of the electrodes was necessary (Figure 3).

Figure 3: modified suspension of electrodes

This system is used by the patient daily as his standard rehabilitation device.

The “7 Degrees of Motion Arm”

This arm is an outcome of the DARPA “Revolutionizing Prosthetics” program. It provides 7 degrees of motion and is controlled by pattern recognition techniques (Figure 4).

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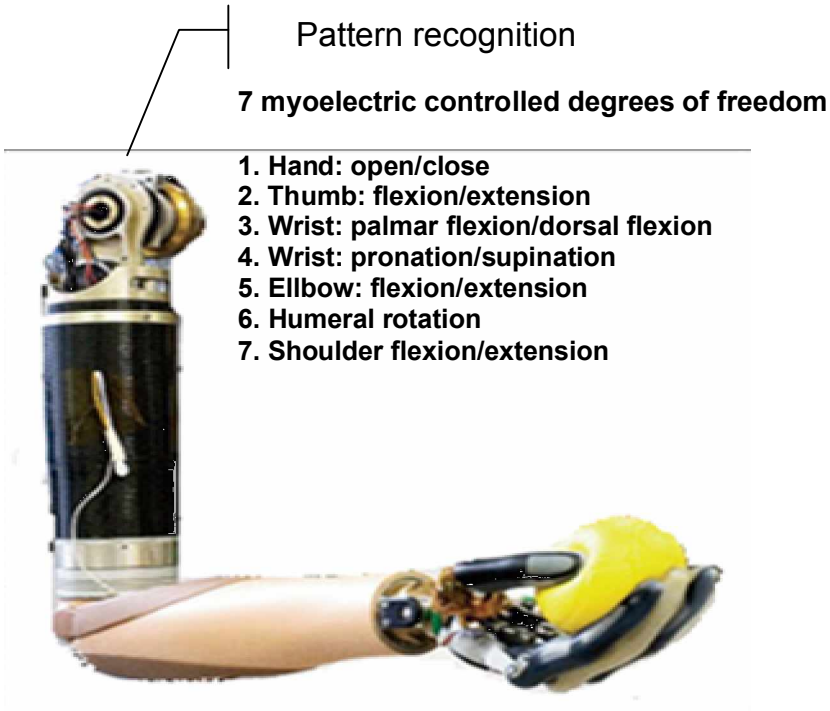

Figure 4: 7 Degrees of Freedom Arm

Up to 24 adhesive electrodes were used for control. This system was used in the laboratory only

Patient outcome:

The patient is using the “Take Home Arm” daily as his standard rehabilitation device. He achieved a sufficient degree of independence and was able to return to work in the logistics and warehouse of a car dealer. The “7 Degrees of Motion Arm” was tested on a regular basis in a laboratory environment. The patient was able to intuitively control all degrees of motion. No additional input devices were necessary.

Outlook

The functionality and the degrees of motion of the “Take Home Arm” will be extended step by step by using the findings of the laboratory tests.

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J Neuophysiol Nov 2007; 5: 2974 – 2982

Abstract

Evolution of the Utah Arm to Improve User Function

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An ambitious re-development of the Utah Arm was guided by direct clinical feedback, producing a list of features to improve the full-electric prostheses.
The goals of the development focused on the following:

- Smoother transition from elbow motion to hand function, and vice versa, i.e., more effortless locking and unlocking by the wearer, with many more locking positions.
- Quieter operation of the elbow, both during the locking and unlocking operations (which can produce an audible "click" of the lock pin}, and during the powered freeswing function, which produces audible motor and gear noise.
- More convenient connection of the prosthesis to the prosthetist's or therapist's computer for adjustment and training, which required a hard-wired connection of cables.
- Wider range of input devices, as well as TDs, to take advantage of all available input sensors and output devices.

Rather than "start over", the technical capabilities and approach of the existing Utah Arm full-electric system were used as a starting point, giving more advantages than disadvantages. Development time is greatly shortened by taking the approach of modifying an existing component, without the risks of using unproven designs. It is important to realize as well – the product is intended for clinical, rather than laboratory use, so practicality is of a very high priority.

Design innovations were directed at improving ease and comfort for high-level prosthesis wearers and improving the ease of fitting. Improvements in the U3 'Plus' system include:

- **Dual Locking System** –a continuous friction lock supplements the high-load lock pin. This innovation provides an infinite number of locking positions for the friction lock. Also, the first-stage friction lock can be more quickly engaged, since the lock pin (requiring motor activation time) only is used with the 2nd-stage of locking. The lock pin, used reliably with all previous versions of the Utah Arm, is activated in the dual-lock system only upon loading the forearm with over five pounds (2.2 kg.). Since the system retains the load-sensing system, the automatic lock required only a modification of the control system to implement the automatic locking. Early clinical use of the U3+ system shows that the 2nd stage lock is activated in ~10% of activities, since high-level prosthesis wearers seldom lift loads of this size.

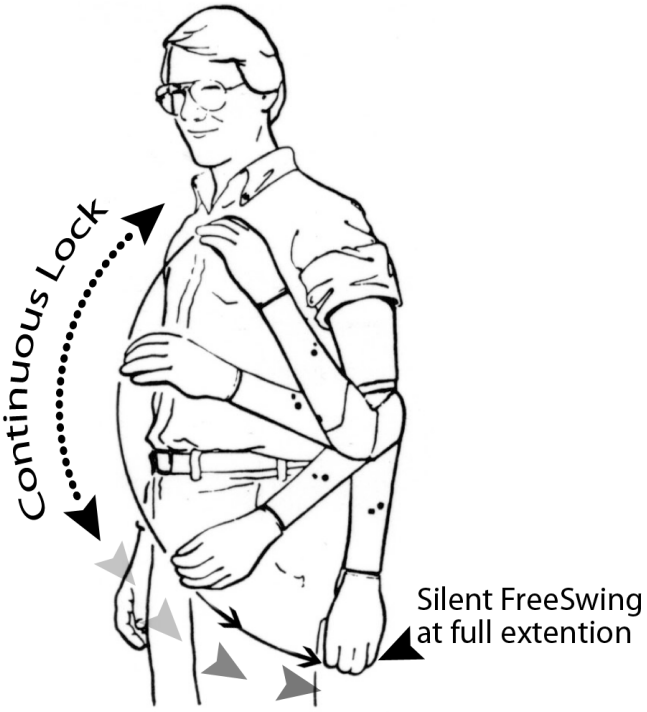

Figure 1 – (left figure) The Dual-Locking system now allows continuous locking at infinite positions. Also, reactivating of flexion or extension is immediate after 1st stage friction locking, since the lock pin is not engaged. 2nd stage locking requires the lock pin, but is utilized only ~10% of tasks, according to wearers. Silent freeswing is now possible, with a new mechanism in the existing Link Arm. The mechanism will automatically disconnect the elbow drive at the fully-extended position of the elbow

• Silent Freeswing effectively lowers the perceived noise of the elbow operation, by incorporating an innovative automatic disconnect of the drive, saving battery power, and eliminating noise. The momentum of the elbow when it 'bumps' slightly at full extension (actually 15 degrees of flexion), disconnects the two-part Link Arm, so that freeswing without any power consumption, nor without sound, is now possible. The mechanism has been laboratory tested to several hundred thousand cycles without failure.

• Wireless Bluetooth Adjustment Communication –an inconvenience during adjustment and/or training during use of the prosthesis has been the hardwire connection of the Computer Interface Module, which limited the activities possible with the prosthesis during adjustment/training. Using the now ubiquitous Bluetooth communication, existing circuit board designs, and existing serial port communication systems can be utilized by connecting a simple "dongle" type transceiver to the prosthesis, and an existing Bluetooth/USB adapter to the computer. Modifications to software are the only other requirements. It is important to note that, the hardwired system, previously used, is still available as a backup whenever the Bluetooth system is unavailable.

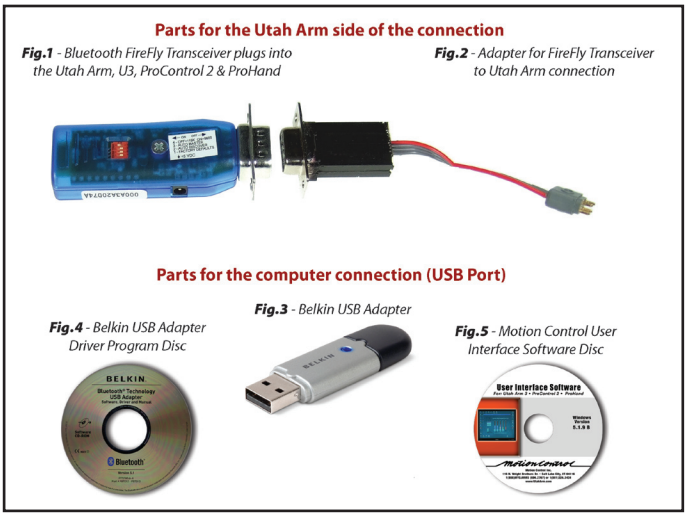

Figure 2 – the Bluetooth adapter system easily converts the existing controller to wireless connection with the clinic's computer, or laptop. A further advantage is that all previous microprocessor controllers made by Motion Control may be used with the Bluetooth system, without modification.

• Compatibility of Inputs and TDs – software innovations allow nearly all input sensors in the marketplace, and the full variety of TDs to be interchangeable with the new system.

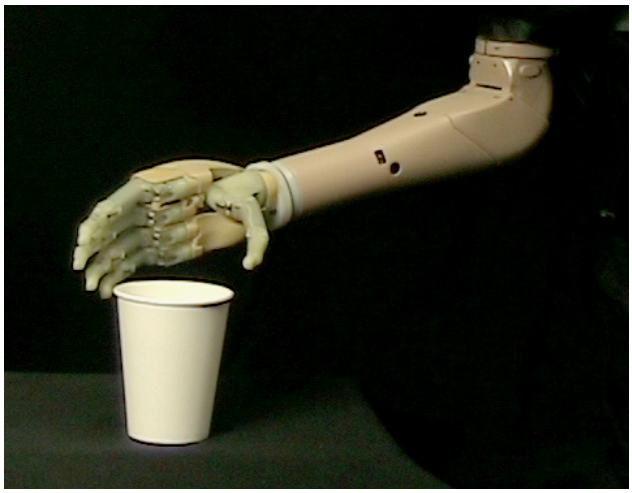

Figure 3 – Compatibility with existing TDs, across the range currently available, is important in prosthetics. No single manufacturer, at present, offers all features available in all terminal devices.

• Modularity – the original design concept has been maintained, providing serviceability important with sophisticated systems.

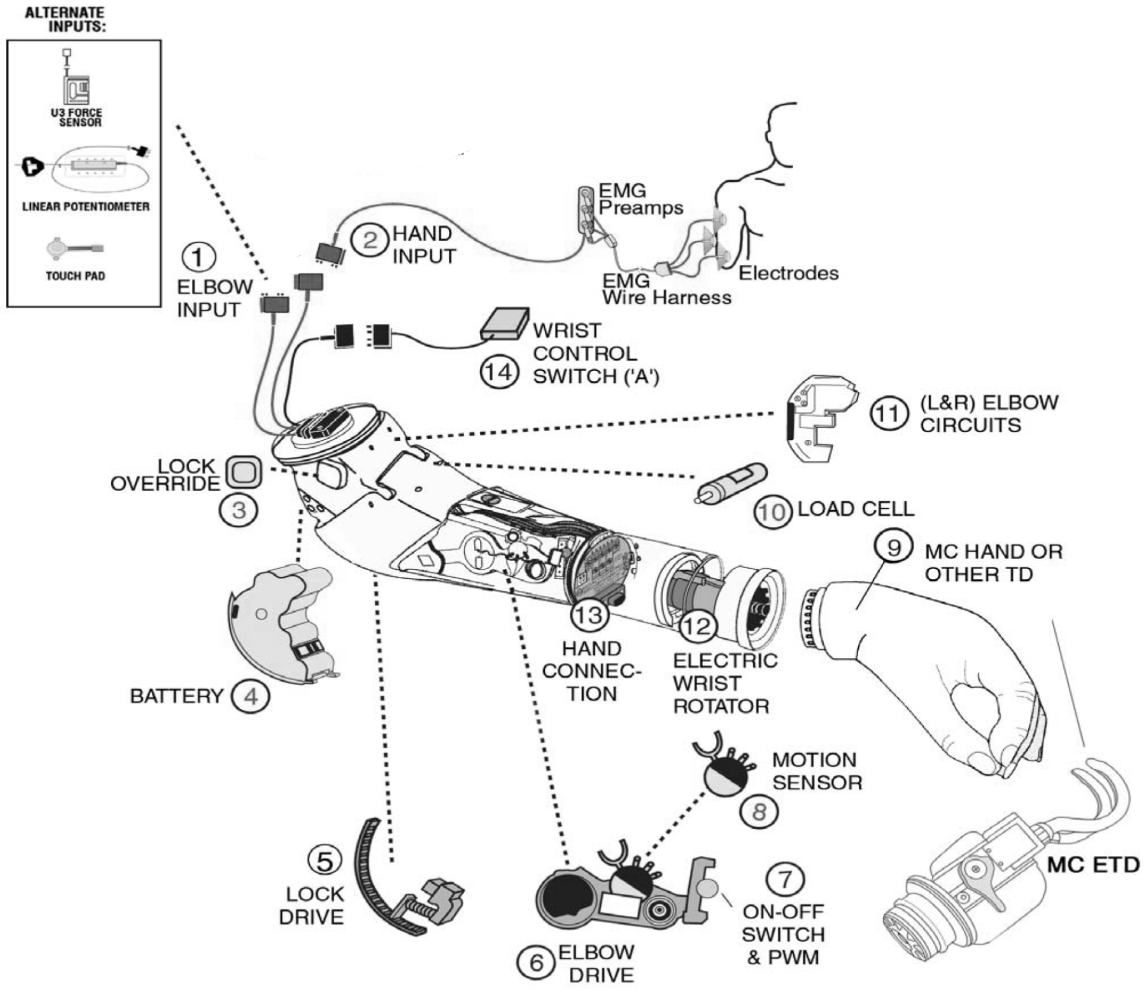

Figure 4 – Modularity has always been a cornerstone of the Utah Arm system. New innovations have been developed without sacrificing this original concept, which allows easier service in the field, and practical interchangeability of components by prosthetists, and in the case of batteries, TDs, and a few other components, by the wearers themselves.

Grip Force Feedback in an Electric Hand - Preliminary Results

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Motion Control, Inc. • 115 N. Wright Bros. Dr. • Salt Lake City, UT 84116

Feedback of sensation has long been the dream of developers (and wearers) of prosthetic hands, and many earlier efforts have made progress, but never a practical commercially-available system. Although grip force feedback (GFF) is an obvious shortcoming in a hand prosthesis, it has been slow to develop because of the innate difficulties of providing consistent and accurate feedback information to the wearer of an electric hand. A truly useful GFF system must provide, 1) true clinical relevance (we feel it should demonstrably improve control of grip force, contribute to a more natural feel, and represent an acceptable ratio of cost to benefits provided), and 2) technically provide a practical system which can operate for months reliably, and be small enough to install into a cosmetic-looking prosthesis.

Early development was conducted at the University of Utah in the 1980's, (Ref.1) resulting in a concept termed: Extended Physiological Taction (EPT) in which the feedback of pinch force was presented to the TD wearer (in laboratory experiments) via a "pusher motor", providing a force against the wearer's skin proportional to the gripping force of the TD. The "mock prosthesis" shown in Figure 1 (left) allowed experiments to be conducted, with non-amputee subjects, to verify the improvement of grip force control by a myoelectric controlled TD, using a TD with GFF. Later, a much smaller "pusher" device (shown at right in Fig. 1) was developed which could practically be installed inside a prosthesis.

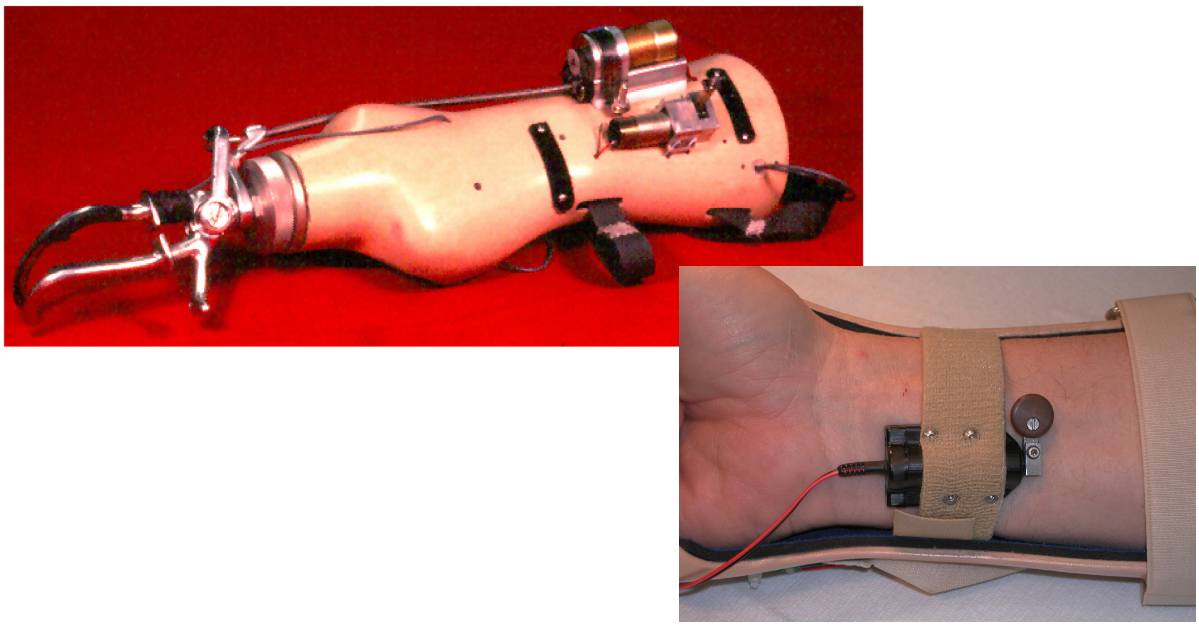


Figure 1 – The concept of Extended Physiological Taction (EPT) was coined to describe the "natural" system of providing GFF to the wearer by means of a pushing device, so that grip force would have an exact correlate to pushing force against the wearer's skin.
Relative simple pushing devices were developed, in initial experiments by Goulding, et al, and ~20 years later at Motion Control, in a much smaller "pusher" device(shown at right). However, a variety of drawbacks were perceived, not the least of which were size and power consumption.

The technical issues requiring solution have significant challenges. Specifically, to yield the practical system presently used, progress was necessary on these topics:

Servo Control to optimize force control for the wearer. A significantly more complex servo control is required to allow the wearer of a GFF system to truly control the force with the precision that the feedback system promises. The drive of the two-speed transmission, found in the MC Hand, is capable of producing very high pinch force, from only a moderate increase in control input by the wearer. This requires a kind of non-linear controller, which allows much higher resolution of the force once the two-speed transition gears down to the higher force gear ratio.

Sensor Development:
Also, as shown in Figure 2, both position and force feedback are utilized in the new servo control, so a position sensor is required in the GFF hand, which had not been part of the MC Hand system heretofore.

Highly rugged force sensing is also required with the MC Hand and ETD devices, for practical GFF in everyday usage. The mounting of the strain gage elements, as well as the routing of wires in a protected manner, among other issues, are all part of the technical challenges which have been refined and tested thoroughly as part of the development process.

Feedback Mechanism
As described, both 'pusher' type feedback and vibratory feedback devices have been explored. The presently used system utilizes a vibration device, which makes a very acceptable compromise between size, energy consumption, low adaptation rate by wearers, and ease of installation. A preliminary installation is shown in Figure 4.

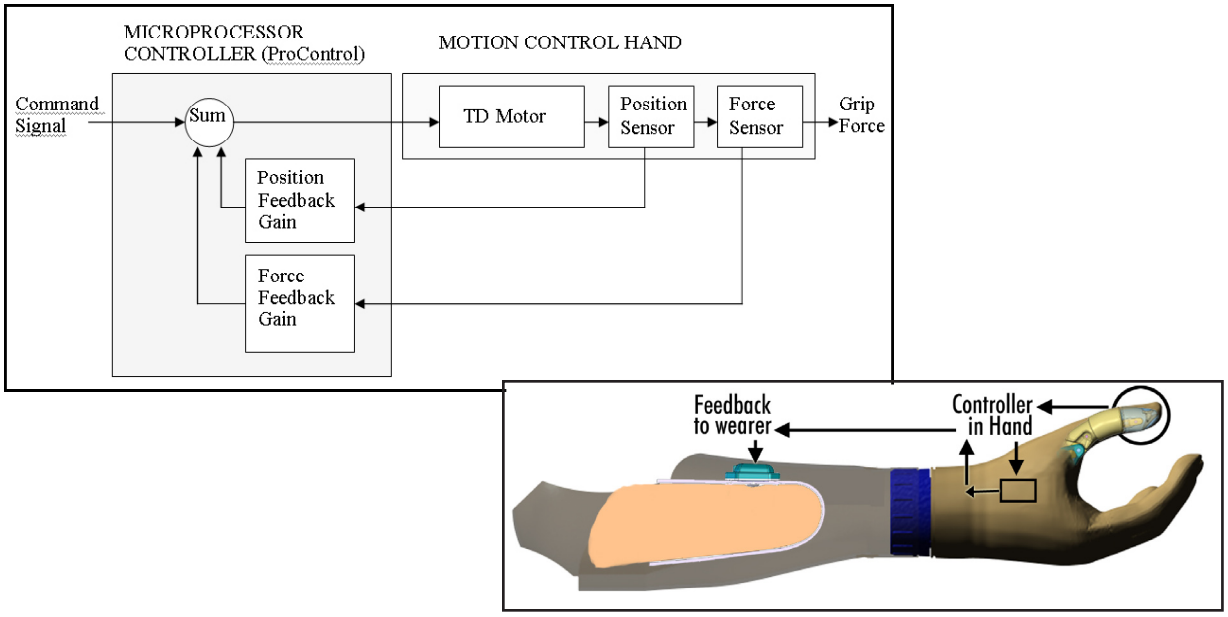


Figure 2 - Technical issues with GFF included evolution of the servo control system (shown simplified on left) which includes both position as well as force feedback. The feedback mechanism (right figure) may be installed wherever appropriate within the socket, and may also be moved to optimize comfort and minimize accommodation.

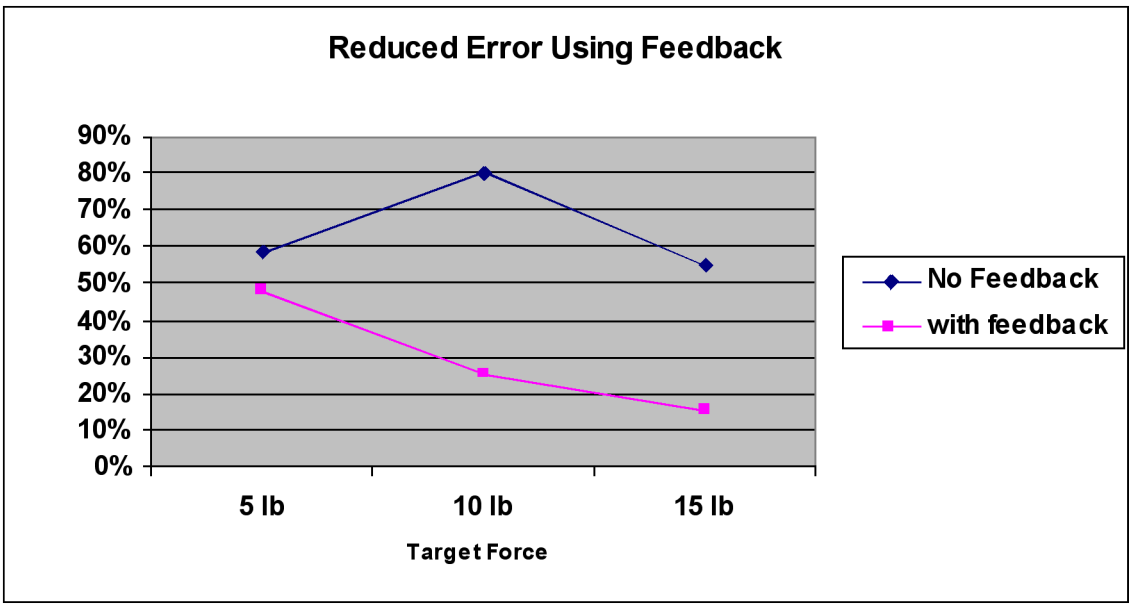


Figure 3 - Results of in-house testing to validate improved control. Tests with non-amputees are considered relevant, because the control method is exactly the same, i.e., dual-channel EMG, and the change of performance attributed to GFF can be presumed to be very similar.

Ongoing Clinical Testing:

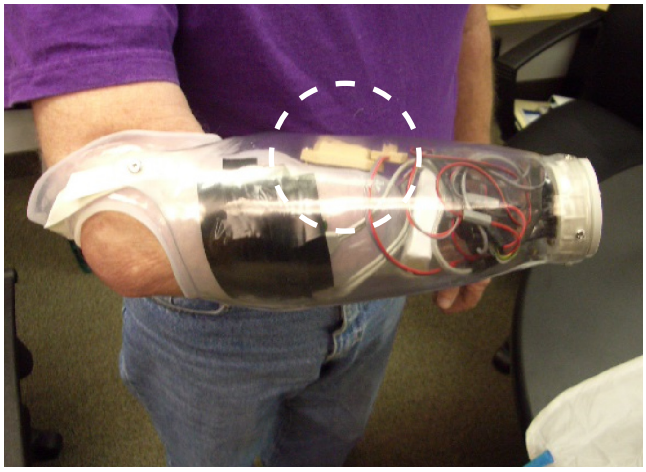
Wearers of these preliminary hands and ETDs, field trialed for over two years, can detect the level of grip force applied by the TD fingers, through vibratory feedback inside the socket. Aside from the sensation of grip force, reported clinical advantages for the wearers include better control of the grip force, as well as safety in the operation of high-strength TDs.

In summary, the feedback experiments showed:

• A small, easily-installed vibrator unit can be placed anywhere the wearer prefers to receive the feedback, and can be moved to new locations, without requiring a new socket fabrication. Field trial fittings indicated preferred placements.

• The grip force may be detected to very low levels of force. Feedback may be selected by the wearer to be at three specific levels of force, or a continuous feedback above a specific level.

Preliminary field trials will indicate the initial contact level of force preferred, as well as the user-selected levels of force at which the wearers choose to have pulsed vibration feedback, or continuous feedback.



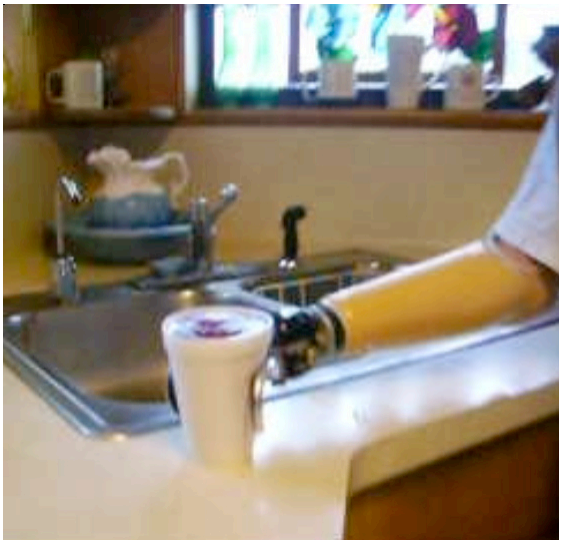


Figure 4 - Actual installation of the feedback element, producing vibration sensed by the wearer's skin, is a process of trial and error, at times. Several locations were used, some only temporarily, in the check socket stage of fitting with this wearer, including the location shown at left. The finished prosthesis, shown at right then could accommodate the vibration device at the preferred location, although changes are still possible if accommodation is experienced by the wearer, resulting in diminished sensation by the wearer. This prosthesis includes the GFF with an Electric Terminal Device (ETD), which also has the advantage of not requiring a glove or hand shell.

Our experience has definitely shown that the availability of grip force feedback in an electric hand prosthesis is effective at improving the wearer's control of grip force, and the sense of a 'more natural' interface between the wearer's and their prosthesis. Definitive and measurable results remain to be performed, beyond this preliminary anecdotal evidence. To date, the system appears to be very promising, and appreciated by the wearers who have had a chance to use it so far.

Reference:

1. Goulding, P.; Extended Physiological Taction, M.S. Thesis, University of Utah. Dept. of Bioengineering, 1984.

DRAFT – Short Paper

Wrist Innovations To Improve Function of Electric Terminal Devices

Harold H. Sears PhD, Edwin Iversen MS, Shawn Archer MS, Tony Jacobs BS

A major challenge in the development of terminal device (TD) and UE prosthetic devices is to add to the functional benefits to the wearer, without greatly increasing the weight, or complexity, or the cost of the prosthesis.

Using existing TD designs, the opportunity existed to increase the function by increasing the degree of freedom available at the wrist, in several ways. Since the existing hands and electric terminal devices (ETD) were both single degree-of-freedom TDs, improving the positionability of the TD can logically improve the gripping orientation and grip security.
Our goal was therefore, to improve positionability of TDs via improved wrist flexion/extension devices, and an improved wrist rotation device, which could be added in a modular fashion to the existing MC Hand, and ETD already developed and used extensively in the field.

Specifically, three wrist components have been developed to increase function of myoelectric TDs:

1. Flexion/Extension Wrist, with manual reposition to three or four positions integrated into both hand and electric hook.

Figure 1 - Wrist flexion in a Hand TD (left) allows flexion to 30 degrees, and extension to 30 degrees. The button for the manual lock engage/disengage may be pushed or bumped on the side of the locking wrist. Wrist flexion in the Electric Terminal Device (ETD) allows a total of four locking positions, at 52 degrees and 30 degrees of flexion, neutral, and 30 degrees of extension.

Evaluation Methods:
Evaluation of the benefits of the Flexion Wrist involved a questionnaire administered by telephone interview. The functional pros and cons were evaluated by rated versus the device used previously by the wearer, so a simple "A vs. B" type comparison could be obtained, with minimal variables. That is, variables are reduced, since the wearer and the prosthesis are nearly

the same for the test device and the comparison device, except for the single component of the flexion wrist. 17 prosthesis wearers were surveyed who had been fitted with the Flexion Wrist in either a Hand or ETD for over two months previously. Some of the most significant ratings obtained are shown below:

RATING OF "USEFULNESS OF FLEXION" (n= 17)

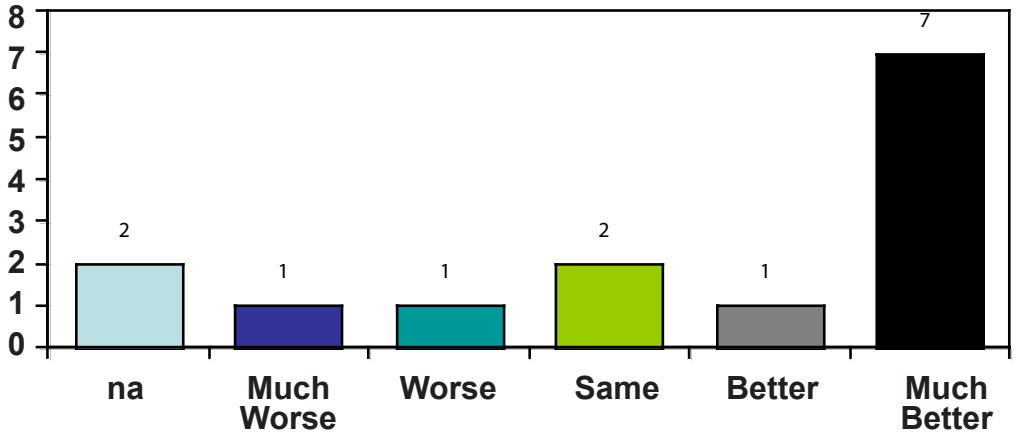
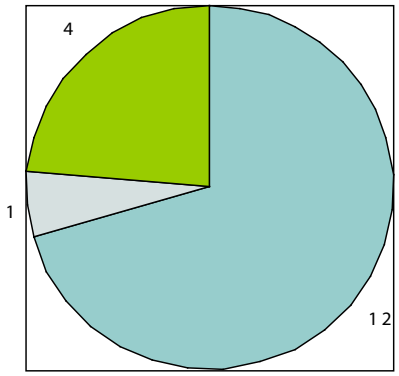


Figure 2 – The 17 patients in the survey were asked to rate the usefulness of flexion in their test prosthesis, compared with their previous prosthesis. The majority (8/17) found the usefulness "better" or "much better" than the prostheses they had used previously, without flexion.

Result #2: DOES THE FLEXION WRIST ALLOW YOU TO:

USE THE PROSTHESIS IN A MORE NATURAL WAY



USE THE PROSTHESIS FOR MORE ACTIVITIES

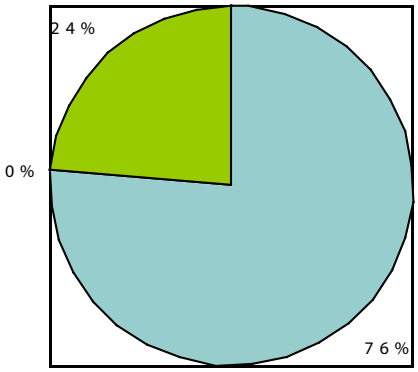


Figure 3 – In another important result, the 17 wearers were asked to simply respond "yes" or "no" to the question, "Do you use your prosthesis in a more natural way?" (left pie chart), and "Do you use your prosthesis for more activities?" (right pie-chart).

2. Multi-Flex Wrist, adds comfort and versatility to the earlier Flexion/Extension joint, by implementing a spring-returned mechanism to the neutral position, while still allowing the wearer to lock the wrist passively in 30 degrees of flexion and extension, as well as the neutral position. The goals of the Multi-Flex device were to reduce reaction forces in the socket, and maintain security of grip on objects that are moved throughout the work space gripped by the TD. The ability to manually lock in three positions is retained, in much the same fashion as with the Flexion/Extension wrist.

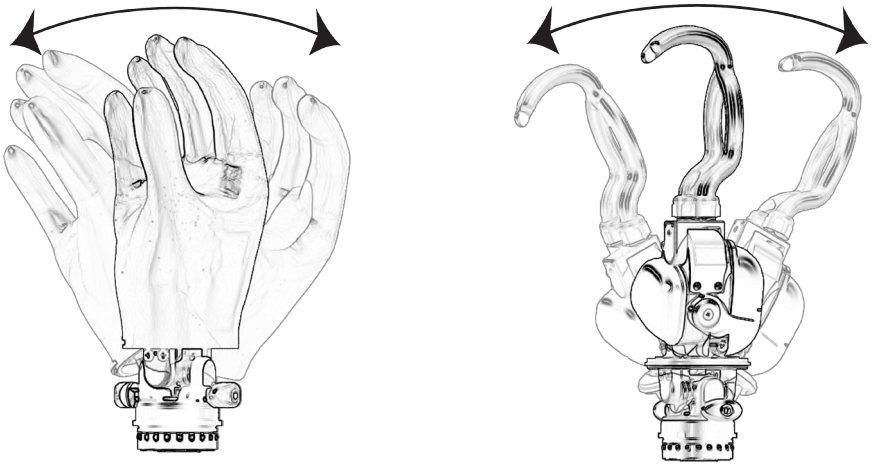


Figure 4 – MultiFlex wrist allows free motion up to 30 degrees of flexion in all directions. Springs return the wrist to the neutral position, and locking is optional at 30 degrees of flexion or extension, and at neutral position. The lock is activated by pushing the locking button.

Data collected for the Multi-Flex Wrist was similar to that collected from Flexion Wrist wearers. Two interesting preliminary results demonstrated that the goals of the development are being met, by showing that wearers can use the Multi-Flex wrist generally unlocked, and that the comfort is generally considered much better.

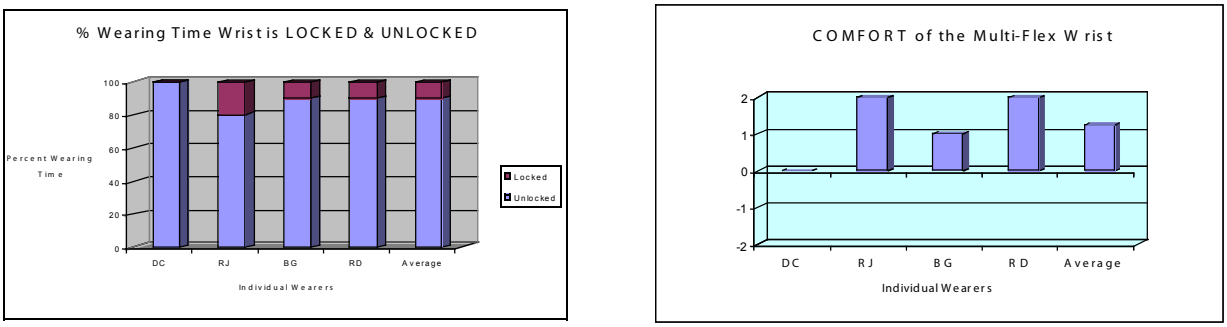


Figure 5- Preliminary results from wearers of the Multi-Flex wrist show that the percent of wearing time that the wrist is locked is quite low (left chart), indicating that the security of grip must be acceptable in most occasions, even with the flexible wrist. Comfort was also rated "better" (+1 rating), or "much better" (+2 rating), by 3 out of 4 wearers surveyed, indicating that reaction forces in the socket are reduced.

3.In-Hand electric wrist rotator.

Powered pronation/supination of the wrist has long been acknowledged as an important contribution to function of the electric TD especially. In earlier an earlier evaluation by the authors (Ref.1), the contribution to function has been documented, but the use of earlier electric rotation devices was nearly completely restricted to pre-positioning of the TD, with a much lesser number of tasks utilizing "active" rotation (functionally rotating the device in the TD, for instance, pouring from a bottle).

Also, earlier rotation devices increase the length of the electric prosthesis, so that anatomical-matching of the length of the prosthesis is not possible (earlier pro/supination devices required ~7.5 cm, in addition to the length of the TD).

Therefore it was felt that a shorter component, which could provide greater torque and speed to the wearer, would increase the functional capabilities of an electric TD.

The device which has been developed by Motion Control is a high-torque and high-speed rotator (3x greater than earlier devices) giving the wearer greater manipulation force in active rotation, and quicker pre-positioning. Since the rotator and hand are built into the same device, both distal to the quick-disconnect wrist, there is much less space occupied in the forearm

Field trials of the InHand Wrist Rotator have begun, but sufficient data has not been collected to this date. However, the few wearers of the device have used it extensively and it appears that it will make a very positive contribution to function for Hand + Wrist electric TD wearers, as well as ETD + Wrist wearers (although).

Figure 6- Electric Hand + Wrist length, compared with Hand alone (left). An example of higher torque of the InHand Wrist – a 1.0kg elbow prosthesis is rotated easily by a wearer, a difficult task with earlier components.

CONCLUSION:

Wrist function is clearly an important contributor to function of the Terminal Device, and thus the prosthesis. When even modest numbers of field trial wearers can be surveyed, quite specific pros and cons can be identified of new these new components.

1. Evaluation of Electric Wrist Rotation…….. Sears and Shaperman, Electric Wrist Rotationin Proportional-Controlled Systems. Journal of Prosthetics and Orthodics, Vol. 10 No.4, Fall 1998.

Abstract submission by Phil Newman of Touch Bionics (technical/clinical presenter TBC)

Multi-articulating Hands & Fingers –

Technical Strategies for Improved Patient Function and Myoelectric Uptake

Since the commercial release of the i-LIMB Hand in September 2007, and the subsequent R&D release of ProDigits for partial hand users, there has been much interest from clinical professionals – to date (18 March 2008), more than 160 patients worldwide have been fitted with the i-LIMB Hand.

This terminal device is the world's first microprocessor controlled, myo-electric hand with proportional control, 4 articulating digits and a thumb which rotates 110 degrees. The i-LIMB hand can support 45 pounds weight and yet the articulating fingers produce realistic and dexterous hand movements that appeal to a wide range of male and female users.

Whilst the technical nature of the hand will always interest the professional – this presentation will address the ways in which technology and traditional thinking are challenged by the experiences and ambitions of patients.

The proposed presentation will dispel myths about grip forces and cosmesis techniques, will address new hand functions delivered by software and will demonstrate the future path for increased patient uptake of myoelectric devices. To date, only one i-LIMB Hand has been returned by a user – many i-LIMB users have previously rejected traditional myoelectric devices.

THE WILMER PASSIVE HAND PROSTHESIS FOR TODDLERS

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ABSTRACT

The WILMER Passive Hand Prosthesis is developed for children aged one through five years. This harnessless hand prosthesis aims primarily at giving the child two arms of equal length. The hand features an easy to control passive prehension function. The movements of the fingers are mechanically coupled to the movements of the thumb. By pressing an object against the fingertips, the hand opens. By slightly tilting the object, it can be grasped. The hand prosthesis can be mounted in a passive friction wrist rotation prosthesis, available in different outer diameters. The hand features a low mass construction, which proved to be very robust and reliable in clinical tests. For the age group mentioned the WILMER Passive Hand Prosthesis is one of the very few available with a prehensile function. It stands out in functionality and in cosmetics as compared to other passive hand prostheses on the market.

INTRODUCTION

For young children, 1 – 5 years of age, with (unilateral) upper limb deficiencies at forearm or upper arm level, the parents sometimes opt for a prosthesis. The reasons can include cosmetics, i.e. the wish to look as complete as possible, or the idea that the use of a prosthesis is beneficial for the development of the child. As an alternative for the myoelectric prosthesis, which is often judged as too heavy, especially for the very young child, we have developed a mechanical, low mass hand prosthesis, Figure 1.

Figure 1.
A young girl wearing a WILMER Passive Hand Prosthesis

METHODS

As point of departure for the design of the hand mechanism, the desire to keep the mechanism as simple as possible was chosen. The hand is of a passive nature. This implies that the other hand is needed to open the prosthesis. This can be seen as a drawback, however, the sound hand is most often already utilized to hold the object and to present it to the prosthetic hand. The advantage is that no harness is needed to operate the hand, as is usually the case in an active mechanical hand prosthesis.

The working principle of the hand is simple, Figure 2. The movement of the fingers is mechanically coupled to the movement of the thumb by a four-bar linkage. An elastic band acts as a closing spring and provides the desired pinch force. By pressing an object against the fingertips the hand opens. Now an object can be tilted in between the thumb and the fingers of the prosthesis. By releasing the fingers, the hand is closed by the spring and the object is clamped between the thumb and the fingers.

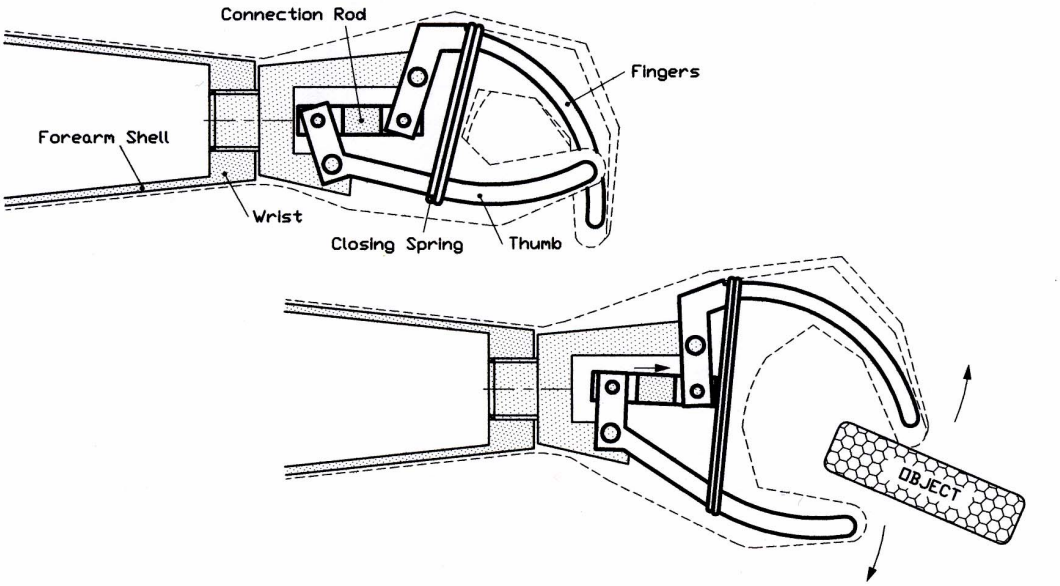

Figure 2.
The mechanism of the WILMER Passive Hand Prosthesis comprises a four-bar linkage. This linkage encompasses the thumb, the fingers, a connection rod, and the hand frame.

To achieve a natural looking operation of the hand mechanism, the plane of movement of the thumb is placed at a 45-degree angle with respect to the plane of movement of the fingers. This implies the need for a spatial four-bar mechanism with an additional degree of freedom. Instead of adding a rotation option along the long axis of the connection rod, some play in the bearings of the connection rod is allowed. This method is preferred as it keeps the mechanism simple.

To couple the hand mechanism to the forearm shell a very simple friction wrist was designed that allows for passive pro- and supination. The amount of friction can be easily adjusted by turning a setscrew.

Through our collaboration with several rehabilitation centers in The Netherlands the

WILMER passive hand prosthesis is supplied to children in the target age group for clinical testing. Over the years almost one hundred of these hands were provided.

RESULTS

The parts of the hand mechanism, and those of the wrist, have been manufactured and the mechanism has been assembled, Figure 3. The operation is as expected.

Figure 3.
The assembled WILMER Passive Hand Prosthesis mounted in the WILMER Friction Wrist.

The technical specifications include a maximum opening width between the thumb and the index finger of 35 mm, an overall length from the wrist to the nail surface of the middle finger of 72 mm, and a total mass of 78 g (without the cosmetic glove).

The passive friction wrist is available in two different diameters: 30 mm and 38 mm, with a mass of 12 g and 20 g respectively. The overall length of the wrist is 13 mm for both diameters.

Some foam material and a cosmetic glove cover the hand mechanism to shape a natural appearance, Figure 4.

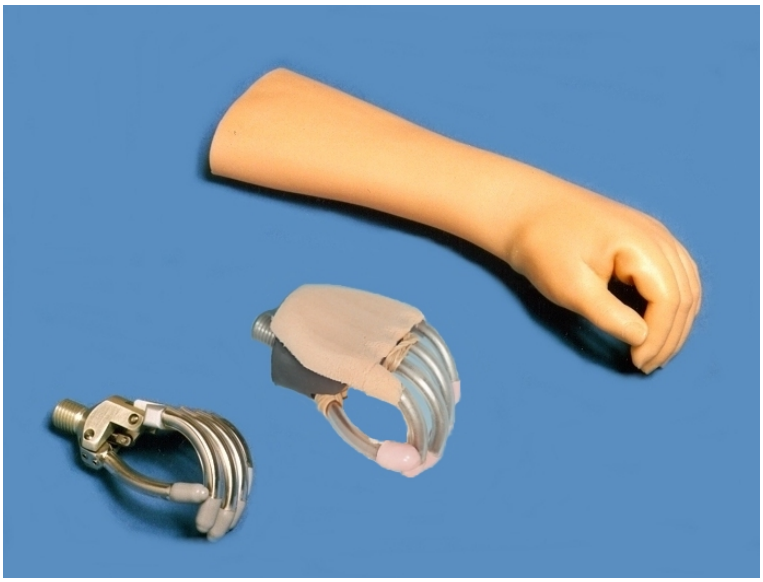

Figure 4.
The hand mechanism is covered by a foam moulding and a cosmetic glove to ensure a natural appearance.

Although a thorough outcome study still has to be performed to determine the functional benefits of this prosthesis, the clinical experience indicates no major problems in the technical properties of the device. The hand mechanism proved to be very robust and reliable. It proved to be insensitive to sand, water and dirt. Repairs are seldom necessary and usually can be attributed to excessive loads, like falls, resulting in deformed fingers or loosened adhesive bonds. Moreover, no special care is needed except for occasional cleaning of excessive debris/sand.

The children and their parents report the hand provides a limited grasping function but excellent cosmetic support for two-armed activities. The prosthesis restores equal arm lengths. The absence of a harness is cherished and especially beneficial for small children, as they do not understand why they need a harness. The hand is used for many different activities like riding a tricycle, and play activities like building blocks, swinging, turning head over heel, etc. Parents report a positive effect on the child's development.

DISCUSSION

The WILMER passive hand prosthesis utilizes a simple four-bar linkage mechanism and is operated in a very simple passive fashion. Therewith the overall mass of the hand is kept very low. The hand offers mainly a support function to the child, which proved very useful for riding a tricycle or in play activities requiring two handed support tasks.

Compared to the only known other passive hand prosthesis for children, the L'iL E-Z hand [1], marketed by TRS Inc., USA, the WILMER Passive Hand Prosthesis provides a more easy grasp of an object as a result of the coupled motion of the thumb and the fingers. In the L'iL E-Z hand grasping is more difficult as only the thumb can be moved. As a result, the other hand is occupied by keeping the thumb open and is not available to present the object to the prosthesis. In the WILMER Passive Hand prosthesis the object is pressed against the fingertips to open the hand. Once open, the object can be tilted in between the fingers and the thumb. Moreover, the WILMER Passive Hand Prosthesis is cosmetically superior, mainly because of the cosmetic glove used.

CONCLUSION

The WILMER Passive Hand Prosthesis is developed for children aged one through five years. This hand prosthesis gives the child two arms of equal length, and an easy to control passive prehension function without the need for a control harness. The hand features a low mass construction, which proved to be very reliable in the clinical tests. The WILMER Passive Hand Prosthesis stands out in functionality and in cosmetics as compared to other passive hand prostheses on the market.

ACNOWLEDGEMENTS

The contributions of the present and former members of the DIPO research group at Delft University of Technology are gratefully acknowledged. We thank our clinical partners of the rehabilitation centres "De Hoogstraat", and "Sint Maartenskliniek" for their co-operation.

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1. L'iL E-Z product information. Available: www.oandp.com/trs. Accessed: 20080618

LEVERAGING LARGER MARKETS TO ENCOURAGE INNOVATION IN PROSTHETICS: DESIGN OF MYOPEN, AN OPEN MYOELECTRIC SIGNAL PROCESSOR FOR USE AS AN INPUT DEVICE FOR VIDEO GAMING AND HOBBY ROBOTICS

Timothy Hanson and Jonathan Kuniholm
Duke University and The Open Prosthetics Project

ABSTRACT

The prosthetic arm market is small; perhaps 70,000 people lack arms in the US. This, coupled with the difficulty of making and controlling a mechatronic hand replacement, has forestalled innovation in arm prostheses in the fifty years since its introduction. Since the Boston Arm project in 1965, the popular press has promised thought-controlled prosthetic arms, yet the promises of scientific research have not often been kept in the clinic. [1] We aim to surmount these obstacles by developing an open myoelectric signal processor targeted at researchers, hobbyists, and video game enthusiasts. Our device is capable of processing 16 channels of surface electromyographic (EMG) (or other data), applying pattern recognition algorithms in real-time via a power-efficient Blackfin Digital Signal Processor (DSP), and delivering the results through ethernet, I2C, RS232, USB, and Lego NXT bus. We hope that this open, commodity level platform will become a disruptive technology, encouraging experimentation in the algorithms and applications of a field that has been sequestered too long in the research lab.

BACKGROUND

Much recent media attention about prosthetic arms has surrounded the Defense Advanced Projects Administration (DARPA) Revolutionizing Prosthetics Program (RP2007/2009) and military amputees from recent conflicts. There remains a disconnect between the reality facing an arm amputee and this coverage in the popular press. As of 2007, there were fewer than 700 US military amputees from Iraq and Afghanistan, of whom fewer than 150 were arm amputees.[2] Despite the relatively small numbers involved, the media coverage may have brought a welcome increase in efforts to improve arm technology. The DARPA prosthetic program has been compared to the Manhattan Project and the race to put a man on the moon. [3] While perhaps an apt comparison in terms of the technical challenges presenting the researchers, the comparison of project budgets in Figure 1 shows that the Revolutionizing Prosthetics Program budgets represents a small fraction of the size of those of the two historical projects. The massive historical efforts in fact compare in size to the R & D costs alone of the current F-35 Joint Strike Fighter program.[4,5,6,7]

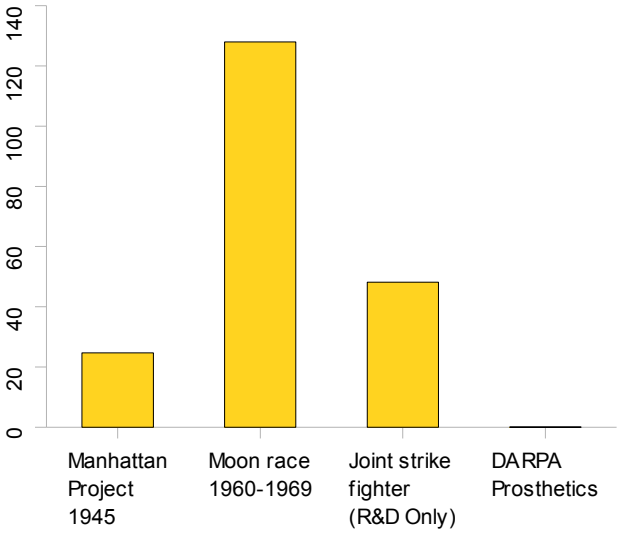

Figure 1: Budgets (Billions of 2006 Dollars)[4-7]

The deficiency that these programs seek to correct is a basic lack of widely adopted advanced upper extremity products. Many amputees, despite being fitted with myoelectrics, return to body powered designs that have changed little in fifty years. Powered products remain slower, heavier, more fragile and more uncomfortable than the alternatives. This is no fault of the companies that serve arm amputees, but the result of a convergence of factors: the small market size, and the difficulties of the problem of replacing an arm. It is not clear that these realities will make the transition of the impressive technology developed by the RP2007/2009 teams to commercial products any easier. It is this "last mile" of the transfer of technology that we have sought to address with this project.

What is it that separates mass markets from underserved ones? The distribution and therefore lowering of costs, in research and development, manufacturing and distribution to a large number of users. It is also the commodification of products in the marketplace, reducing profit margins because of the availability of alternatives. As illustrated in Figure 2, the buying power in the current market for powered arms cannot hope to challenge that for razors and blades. [8,9,10,11] The ability of these larger markets to overcome higher fixed cost barriers to entry explain the pace of innovation in those areas.

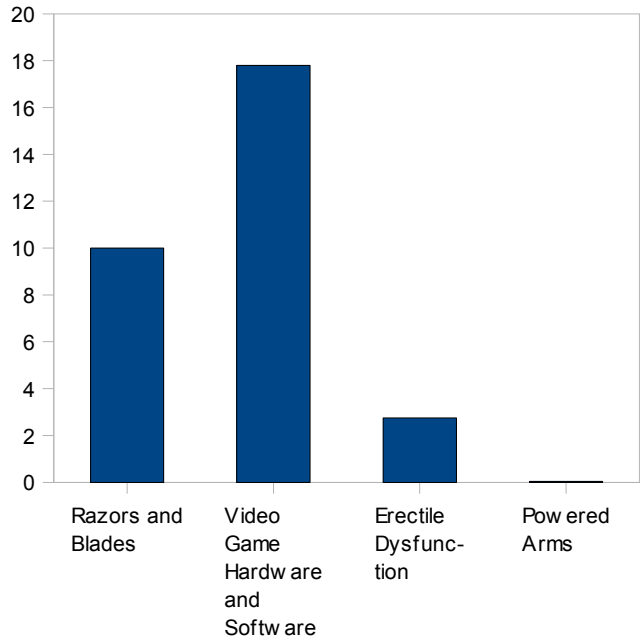

Figure 2: Recent Annual Sales ($Billions)[8-11]

PARASITIZING A LARGER MARKET TO LEVERAGE TECHNOLOGY

Pattern Recognition

While other avenues exist for overcoming these barriers, the one that we have chosen is to seek application in markets with fewer limitations in buying power for underused technology in which we are interested. One of the most promising possibilities for prosthetic arm control is myoelectric pattern recognition. Despite more than 20 years of research in this area, and the advent of prosthetic hands with kinematic capabilities exceeding the possibilities for conventional control, no commercial product in arm prosthetics uses it. In order to help make this or any other advanced signal processing technology more accessible to amputee users (and manufacturer and prosthetist users), we have created a product that makes it possible to use pattern recognition or any other signal processing algorithm on skin surface EMG signals for a variety of applications.

In support of the project, a Matlab code base has been generously provided by Dr. Kevin Englehart at the University of New Brunswick (UNB), in Fredericton, Canada. The software, named CEVEN, contains code written by Dr. Englehart and his students, and was assembled into a single package by Blair Lock. This package provides training, classification of EMG signals through a variety of methods, and visual simulation of arm movements that can be paired with pattern recognition. UNB provides no support for the software, and all support and discussion should be maintained through the Google Code site: http://code.google.com/p/myopen/. Ultimately, the project seeks to port the code to a python or similar high level and free platform, but the Matlab version is available in the meantime.

Using the Power of the Video Game Market

Of the collection of large mass markets shown in Figure 2, that for video games and game hardware is the largest. Worldwide spending is almost $18 billion on items for this type of entertainment. The industry has recently undergone several transformations in human interface, from the use of wireless and accelerometer technology in the Wii controller, to the Emotiv controller's use of EEG signals for “brain control”. Pattern recognition is an obvious very capable addition to the battery of ways in which humans can intuitively interact with a virtual world.

Openness Ensures Access to Technology

To ensure that all stakeholders can actually take advantage of the benefits that come from the use of pattern recognition in the video game market, we have chosen to release both our hardware and software designs under the open source GNU Public License. By ensuring that the designs will always be freely available and changeable, we can break down the silos that separate different uses for the technology, allowing user customization for purposes that we have not yet envisioned.

This approach has been around for a while in software, but is relatively new in hardware. Several current devices use it, including the Neuros OSD

video device, and the Buglabs BUG. The multi-purpose linux BUG device with swappable and easily configurable components includes a GPS, touch screen and motion sensor, with which users can make their own embedded devices. Because of the added complexities involved in dealing with the physical world, this approach is still very much the beginning of an experiment.

TECHNICAL DETAILS

Our first iteration of the MyOpen EMG signal processor is based on the Blackfin DSP and has a number of features to encourage development, customization, and interfacing to the greater world. The board for this device is effectively broken into two parts: the digital section, including the DSP and all communication ports, and the analog section. The two sections are electrically isolated for safety.

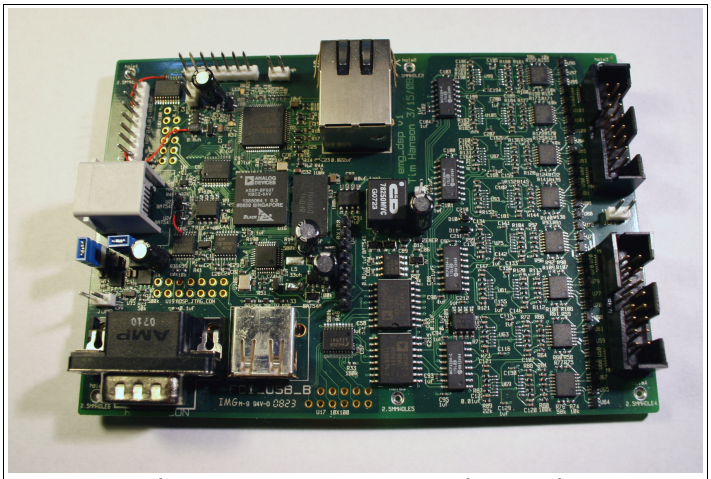

Figure 3: The MyOpen EMG Digital Signal Processor

The analog section consists of 16 channels of differential EMG amplification. At the head of this signal chain is a low-cost, low power instrumentation amplifier(INA2322) followed by a high pass (fc = 1hz) and a low-pass (fc = 400hz) stage. Both filtering stages are in a multiple-feedback configuration, and both provide a gain of 4; the instrumentation amplifier provides a gain of 30 for a total system gain of about 480 (54dB). The filtered EMG signal is digitized at a resolution of 13 bits with the four-channel MCP3304, and this data is sent to the digital section of the board via ADUM2400 magnetic isolators. The analog section has a 0/2.5/5v split-rail powered via a MAX253, transformer, and low-noise regulator.

The digital section features the DSP, 32Mbytes SDRAM, ethernet PHY & magjack, RS232, USB, JTAG, and Lego NXT compatible I2C / TWI. The board also features a small 128x128 pixel serial LCD for debugging and user interface, as well as header breakouts for 8 general-purpose pins on the DSP. Each of these devices may be powered down to conserve battery power. Furthermore, a low-power pin-compatible Blackfin BF534 chip may be substituted in for the faster BF537 when the ethernet interface is not required. The core voltage for the Blackfin may be runtime switched between 0.8V and 1.2V, again to save power. This and the system 3.2V bus are supplied through two efficient synchronous buck converters from either a 3.7V lithium cell or a 5V external supply.

The first revision of the board is 6 layers, routed on 0.005" trace/space rules, with 0.006" minimum via hole under the DSP/SDRAM, and is 3.3"x4" in dimensions. The board can be made smaller in the future by placing components on both sides.

ACKNOWLEDGEMENTS

This research was supported by the Open Prosthetics Project, specifically with generous donations from Douglas Abrams and William Becker. The work was indulged by Dr. Rob Clark at Duke University, and by the DARPA Revolutionizing Prosthetics Program, DARPA contract N66001-06-C8005, and by NSF grant DGE-0221632.

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APPLICATION INDEPENDENT ASSESSMENT OF CAPACITY FOR MYOELECTRIC CONTROL

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INTRODUCTION

There is a growing need for objective and quantitative tools to measure the outcome of assistive technologies (AT), such as powered prostheses and communication aides for people with disabilities [1,2]. The most typical types of assessment are task completion tests or performance tests, where the user performs specific tasks relating to practical daily activities and the time required for task completion is used as the assessment.

The main purposes for such outcome measurements are the assessment of patient improvements, the selection of appropriate AT devices for patients and the evaluation of newly developed AT devices. The appropriate choice of an assessment method requires careful consideration, because assessment results can be influenced by many factors. For instance, applications to AT device selection, where a therapist would investigate the outcome differences for different AT devices; assessment results may be influenced primarily by task proficiency, which thus makes the appropriate selection of an AT device difficult.

This paper introduces a basic form of assessment that is both task- and application-independent, which focuses on the operation capacity of an input device to an AT. In most AT devices, single or multiple switches are widely accepted as the de facto standard input devices, so the proposed method evaluates the operation capabilities of single switches.

At the same time, there are also a number of research projects that are seeking to develop augmentative and alternative input methods for AT devices. However, because of the wide variety of impairments, certain restrictions still remain on increasing the number of candidate input methods. Accordingly, we are working on general purpose myoelectric interfaces that are compatible with various commercial AT devices. Preliminary evaluations of the myoelectric interface are also presented in this report.

METHODS

General-purpose myoelectric switch interface module (EMG Switch)

Figure 1 is a photograph of the prototype switch interface module, which has one input port, three output ports, two volume adjustments and three LED indicators for each output port. The input port receives the sensor outputs, which are integrated myoelectric signals ranging from 0-5V. The two volume adjustments are used to adjust amplifier gain and threshold, which are described below.

The specifications for the output ports are illustrated in Figure 2. The switch module has two thresholds to determine the output signals. Threshold-1 is adjusted using the volume in Figure 1 (threshold adjustment) and Threshold-2 is fixed (4.7V). Output-1 and Output-2 maintain ON states when the input signal exceeds Threshold-1 and Threshold-2, respectively. Output-3 turns to an ON state when the input signal exceeds Threshold-2 and maintains an ON state while the input signal is above Thresold-1. Users can maintain an ON state with low myoelectric signal intensities using Output-3.

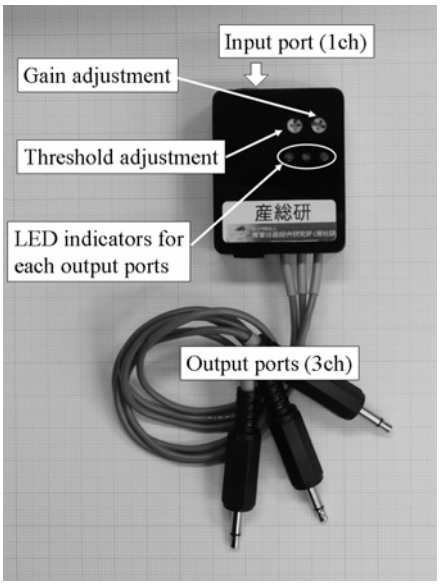

Figure1: Prototype switch module.

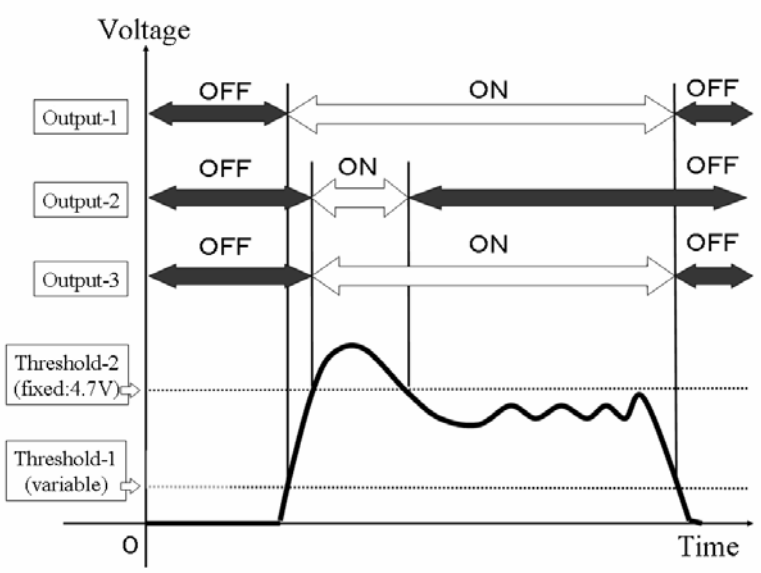

Figure 2: Switch output specification.

Application independent assessment

Eleven essential items for simple switch assessments are reported in [3]. These items were identified by a focus group meeting with six occupational therapists. We selected five items which relate to the operation capability of single switches: namely, (1) “Switch Close Delay”, (2) “Switch Open Delay”, (3) “Switch Endurance Period”, (4) “Switch Repetition Time”, and (5) “Switch Timing Spread”.

(1) “Switch Close Delay” and (2) “Switch Open Delay” refer to response times, as the time elapsed between stimuli and the switch open/close operations. (3) “Switch Endurance Period” refers to period during which a user is able to maintain switch closure. (4) “Switch Repetition Time” relates to the speed of switch operations.

(5) “Switch Timing Spread” relates to the ability to use scanning methods that enable single switch users to select elements from some candidates, such as characters from a virtual keyboard or options from a menu. One of the elements is highlighted, and the highlight moves from one element to the next at pre-set times. The user, therefore, requires the capability to operate a switch in synchronization with the highlight movement. This “Switch Timing Spread” assesses whether response times are premature or delayed, where a premature response indicates the time prior to the target timing and a delayed response indicates the time lag after the target.

Assessment tool

A prototype assessment software tool was designed to be used on a personal computer (PC). Figure 3 shows a screen shot of the tool. There is a target cross-shape (small, light-color) on the center. The operator is asked to move another cross-shape (large, dark-color) to overlap the centers of the two cross-shapes. The operation procedure is similar to that of a claw vending machine or toy crane machine. First, the horizontal movement is activated with a switch operation. Once the horizontal movement is terminated, it can not be activated again. Then, the operator deals with the vertical alignment. After the vertical motion, in the case of a claw vending machine, a crane and a claw are activated, but in the case of this tool, window zooming is activated. The tool window can be zoomed by repeating the switch operations.

Figure 4 shows another assessment mode for the tool, which is referred to as a BLIND mode, in contrast to the BASIC mode in Figure 3. Half of the window is shaded in order to hide the cross-shape. Movement of the large cross-shape is visible in the BASIC mode; therefore we can predict appropriate timings for switch operations. The purpose of the BLIND mode is to remove prediction effects in order to evaluate the response times of the switch operations.

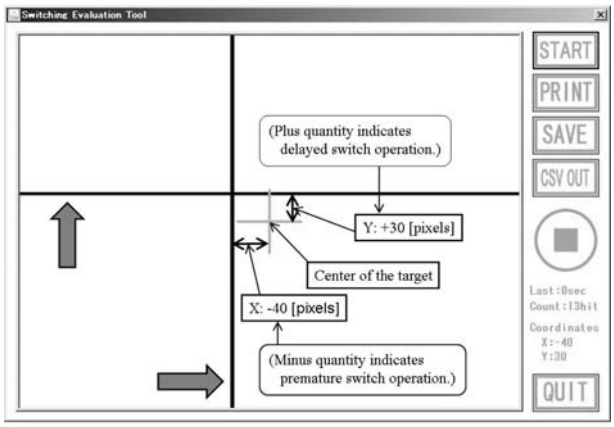

Figure3: Screen shot (BASIC mode)

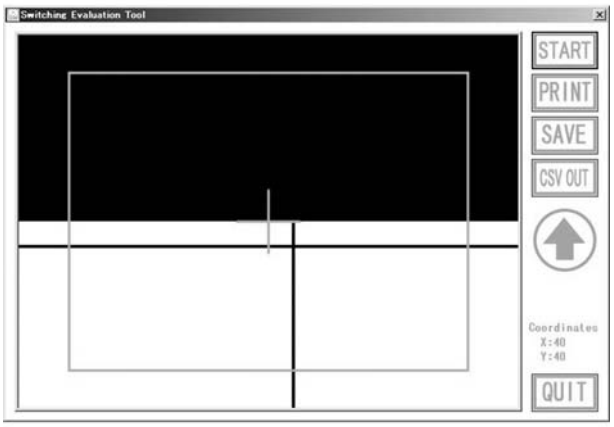

Figure4: Screen shot (BLIND mode)

There are two different switch activation modes; a direct mode and a toggle mode. The function of the direct mode switch is the same as a simple mechanical switch, where the output signal is maintained in an ON state during switch closure. In contrast, the output signal from the toggle mode switch toggles back and forth between the ON and OFF states.

Table 1 summarizes single switch operation capabilities that are assessed with the prototype tool. It should be noted that, for the current software version (ver.1.1), the switch operation capabilities (1), (2) and (5) were assessed in terms of the distances between the two cross-form centers, whereas the result for (4) is the number of switch repetitions. Capability (3) is not yet quantitatively evaluated with this version.

Table 1: Single switch operation capabilities assessed by the tool.

	Direct switch mode	Toggle switch mode
BASIC mode	(3) “Switch Endurance Period”	(5) “Switch Timing Spread”
BLIND mode	(1) “Switch Open Delay”	(2) “Switch Close Delay”
Window zooming	(4) “Switch Repetition Time”	-----

Experimental setup

The experimental setup for evaluations with the proposed assessment method is as follows. As this is a preliminary evaluation, one healthy man participated in this experiment. The myoelectric switch interface (referred to as EMG switch) and a variable pressure switch (joggle switch; Esterline Corporation) were used. The joggle switch can adjust the actuation force by rotating the switch-cover from 200 grams to 1,500 grams. We used 1,500 grams (refer to switch A) and 200 grams (refer to switch O) pressure settings for the experiments in order to evaluate operation capability with different activation pressures. These switches were connected with the PC using a USB switch interface box (Nandemo Switch USB; Techno-tool Corporation).

RESULTS

Figures 5-8 presents the results of one assessment (average ± standard errors; 100 replications) for "Switch Close Delay", "Switch Open Delay", "Switch Repetition Time" and "Switch Timing Spread", respectively. These results suggest that the differences among different kinds of switches can be assessed with this method.

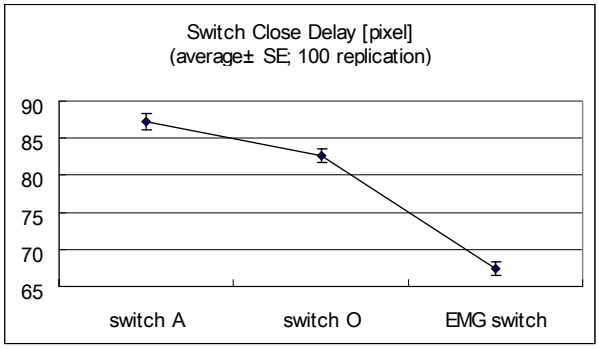

Figure 5: Switch Close Delay.

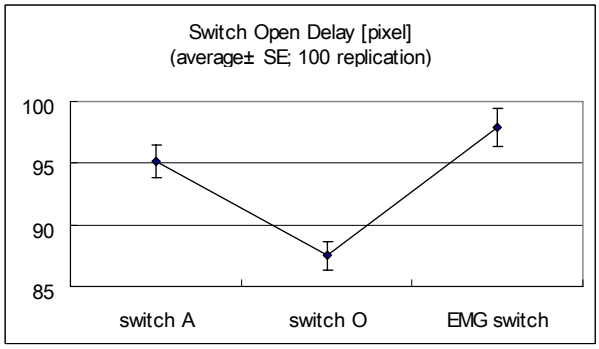

Figure 6: Switch Open Delay.

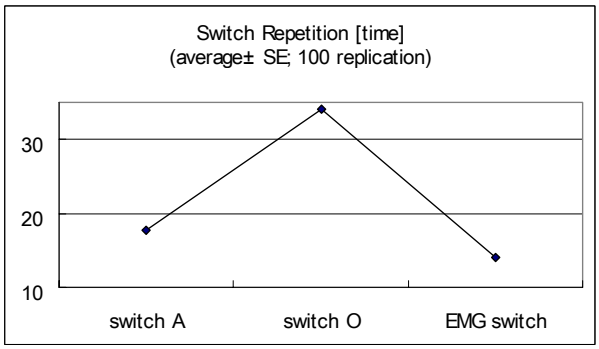

Figure 7: Switch Repetition Time.

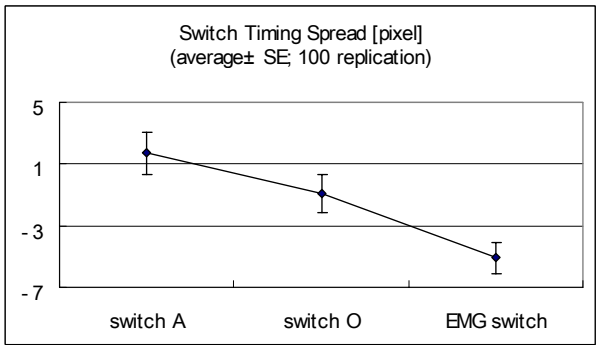

Figure 8: Switch Timing Spread.

DISCUSSION

This paper has only addressed the differences among types of switches, but assessment results differ due to different operation skills of people, between AT devices, and due to environmental conditions. This method can be applied to evaluations of such factors. Socket fitting for myoelectric prostheses may also be assessed with this method.

A general-purpose myoelectric switch interface was also introduced in this report. We believe that clinical information concerning various applications of the myoelectric interface can contribute in the wider application of powered hand prostheses.

The relationships between application independent and application dependent assessments must be examined in future works.

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DESIGN AND PRELIMINARY EXPERIENCE WITH FLUIDHAND MK III

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Abstract

The third generation of the fluidic hand currently is in the process of clinical trial. The new hand combines functionality and cosmesis. It has the natural shape of a human hand and multiple new functions. The grasping operation is built in an adaptive way according to the biological example. This means that objects are enclosed with positive locking. The big contact area and soft passive elements considerably reduce the grasping force required to hold an object securely. Starting from a natural basic hand position, all actuators can be controlled individually by two myoelectrodes in the socket. The software allows for a quick selection of the most important grasping patterns. Besides the multifunctional control, features of the new hand like the force feedback system which gives the patient a sense of feeling and operation of a (computer) keyboard with the index will be depicted.

INTRODUCTION

After upper limb amputation, a basic part of the rehabilitation process for many patients is a vocational retraining from skilled manual work to office work, including the use of computers. Additionally, the use of electronic devices like computers and game consoles is playing a growing role in the adolescents' recreational behavior [1]. A survey of the use of computer keyboards by adolescent prosthetic hand users revealed that at the age of 11 to 15 years a computer is used up to 180 min on an average day with a mean value of 77 min [2]. The prosthesis is used by 5 out of 11 (45%) adolescents for typing on the keyboard and if they do so, they rotate the hand by 180° in order to type with the tip of the thumb. From the interviews, it is reported that this type of keyboard writing is exhausting. The desire of prosthetic hand users for an improved functionality was expressed in several surveys [3,4].

TYPING WITH THE INDEX

The new hydraulically driven prosthesis is a further development of our hands presented in [5]. It is made of a carbon-fiber composite and a high-strength aluminum alloy. The index finger, middle finger, and thumb can be operated independently. This allows for two different options to operate a keyboard with the new prosthesis (Fig. 1a and b).


Fig. 1a and 1b: Options to operate a keyboard with the new prosthesis. Either the index is extended and the other fingers flexed, or vice versa.

FORCE FEEDBACK

An important feature to implement when designing a prosthetic hand with increased functionality is a force feedback system to the user. We have chosen a small vibrator unit to transmit information to the wearer of our electric hand [6]. As a result, different levels of grip force applied can be detected and allows for better control of the grip force. The levels of feedback to the wearer are applied at user-selected levels of force. In order to avoid adaptation to the signal the vibration stops automatically after a predefined time and only restarts when a different level of grip force is applied.

MULTIFUNCTIONAL CONTROL

The aim of the development was to have a robust, simple-to-learn control of multiple grasping patterns of low complexity for the user. Unlike other research groups [7], we decided to use only two standard surface myoelectrodes (Myobock 13E125) placed over the flexor and extensor muscles of the residual limb. During a short period of training (about 15-30 minutes), the sensors were scanned with a frequency of 100Hz and each subject generated 10-15 repetitions of 5-8 different replicable natural movements (virtual movements). Each movement represented a different switch signal for a grasping pattern. To classify the signals, a new classification scheme was applied. All subjects were able to operate the automatically adapted system with a signal reproduction reliability (classification rate) of 90.4 to 99.5 per cent [8]. Hence, a more physiological control was introduced, as the switch signals do not have to be learned by the individual, but are taught to the control software that automatically adapts to the abilities of the individual. First experiments that evaluate the control system in repeated sessions proved that training leads to even increased classification rates and shorter switch durations. It was also shown that small variations of the electrodes placement (i.e. by taking off and putting on the prosthesis again) do not lead to significantly different results, if two training sessions are used for classifier training [9]. Examples of different switch signals are depicted in Fig. 2 a-c.

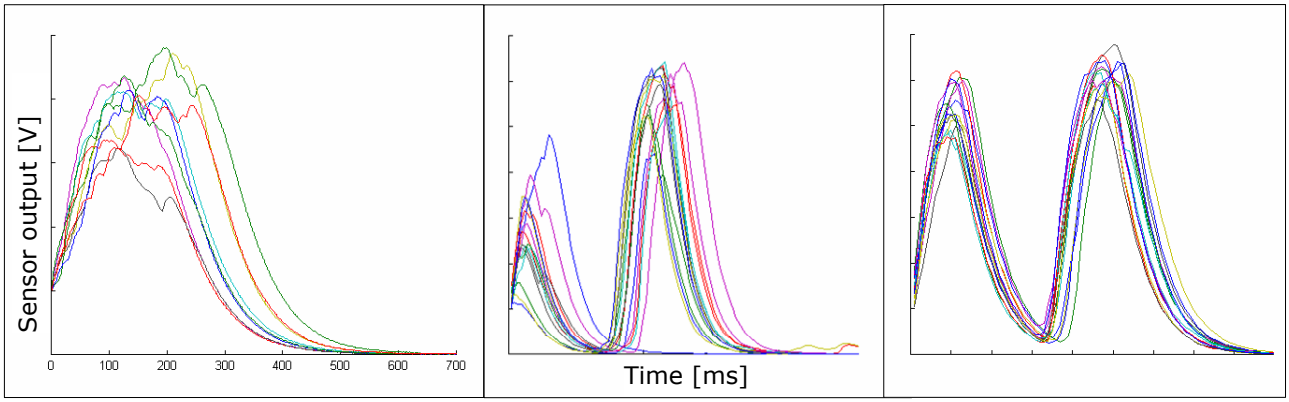

Fig. 2a-c: Three different examples of switch signals.

ACKNOWLEDGMENT

This work received major support from the limb fitting center of the Heidelberg Orthopaedic University Hospital, Pohlig Orthopädietechnik, Traunstein, Germany, and Brillinger Orthopädietechnik, Tübingen, Germany.

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THE EFFECTS OF ELECTRODE IMPLANTATION AND TARGETING ON PATTERN CLASSIFICATION ACCURACY FOR PROSTHESIS CONTROL

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INTRODUCTION AND BACKGROUND: Many researchers have attempted to recognize patterns of muscle activity associated with different movements of the phantom limb and link these patterns to movements of the prosthesis. Researchers have examined a variety of different classifiers and extracted complex features from the electromyographic (EMG) signals to maximize classification accuracy. However, nearly all of these efforts used surface electrodes.

Surface electrodes are advantageous because they are cheap, non-invasive and have a large pickup area. Extracting features from these recordings can allow the classifier to parse out the activity from the different muscles that sum together to produce the myoelectric signal and may increase the information available to the classifier.

Alternatively, intramuscular electrodes may be advantageous for multifunctional prosthesis control because they record focally from deep muscles, provide consistent recording sites as the user changes arm orientation or dons and doffs the prosthesis and reduce crosstalk. However, only two groups have investigated intramuscular EMG for pattern recognition based control [1-4] and only Hargrove, et al. [1] compared surface and intramuscular electrodes, recording from sixteen untargeted surface and six targeted intramuscular channels.

As well as almost solely utilizing surface electrodes, previous studies in pattern recognition-based multifunctional prosthesis control have either targeted the electrodes to specific muscles or used untargeted electrode arrays. However, no previous work has attempted to determine which approach is superior by directly comparing targeted and untargeted electrodes.

Untargeted electrodes are simpler to implement and are preferable for both intramuscular and surface recordings. Socket fabrication can be simplified if the surface electrodes only need to be arranged in an array instead of targeted to specific muscles. Additionally, targeting implantable sensors (such as the IMES [5]) to specific muscles is not a trivial task and would likely require approaches such as ultrasound guidance to properly orient the implants in specific muscle bellies.

Given that the effect of either electrode targeting or electrode implantation has rarely been examined, the goals of this work were to compare the classification accuracies of multifunctional prosthesis classifiers that use either surface or intramuscular EMG as well as those that use either targeted and untargeted electrodes. Further details are available in Farrell and Weir [6].

METHODS : The study investigated four different electrode conditions; targeted surface (TS) and untargeted intramuscular (UI) recordings were collected in one half of the experiment while targeted intramuscular (TI) and untargeted surface (US) data were collected in the other half. Eight proximal forearm muscles were chosen for the targeted electrode sites (Fig. 1).

The untargeted surface (US) and untargeted intramuscular (UI) electrode arrays were equally spaced around the forearm. The insertion depth of the untargeted intramuscular electrodes was varied based upon the size of the subject's forearm and ranged from 1.5 to 2.0 cm. Closely spaced fine wire electrodes were used to acquire the intramuscular signals.

The targeted surface (TS) sites were located by finding the site that produced the largest amplitude signal as the subjects repeatedly produced movements specific to each targeted muscle

(e.g., pronation was the test movement for pronator teres). The targeted intramuscular (TI) sites were primarily found via palpation and prior experience. Proper location of the intramuscular electrodes was verified by electrically stimulating the muscle and observing the induced movement.

Each of the eleven subjects (mean age 29 +/- 8 years; 7 males, 4 females) performed a series of trials in which they alternated between relaxing and performing one of twelve movements of the hand and wrist (Fig. 2). Four contractions were contained in each trial and four trails were collected for each movement. A four-fold cross-validation technique was used to increase the robustness of the classification accuracy estimate.

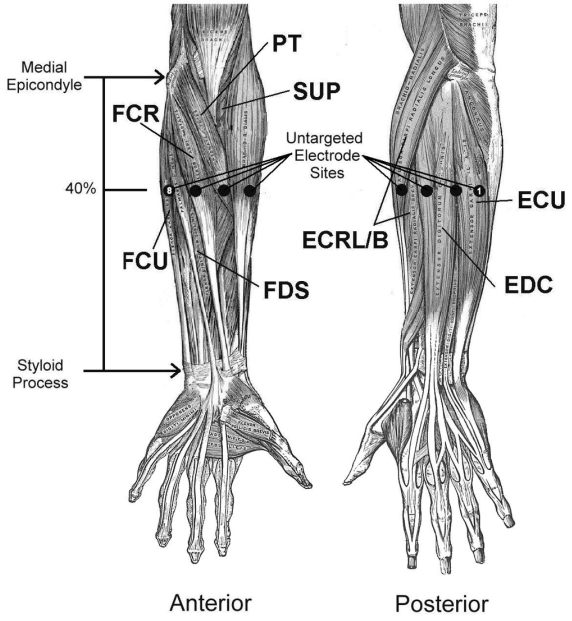

Figure 1 - Electrode sites: Targeted electrodes were located in/over the muscles shown and the untargeted electrodes (large black dots) were equally spaced around the forearm.

Time domain features (TD), auto-regressive coefficients (AR) and root-mean-square (RMS) data were extracted from each analysis window. Classification accuracy is dependent upon a number of different variables. Farrell [7] optimized each of these variables for the data collected in this study before comparing the effects of the different electrode conditions. The reported accuracies were obtained from a state-based linear discriminant analysis (LDA) classifier with overlapped 160 ms analysis windows (no majority voting) and training data that included the onset transient of each contraction.

Accuracies were calculated for feature sets that both included and excluded additional signal features (TD+AR). A two-factor, two-level (targeted/untargeted and surface/ intramuscular) repeated-measures ANOVA with Bonferroni correction was conducted on each data set.

Twelve different movement classes were chosen to make the classification problem difficult in an attempt to highlight the potential differences between the electrode conditions. However, current commercially available devices are not capable of producing the full set of twelve movements. Therefore, classification accuracies were calculated for smaller numbers of output classes to examine the potential differences in the electrode conditions for more currently clinically relevant problems. A two-factor (electrode type and number of classes) repeated-measures ANOVA was conducted on the resulting classification accuracies. The subsets were: 2, hand open/close; 4, hand open/close + wrist pronate/supinate; 6, hand open/close + wrist pronate/supinate + wrist flex/extend; and 8, hand open/close + all wrist movements.

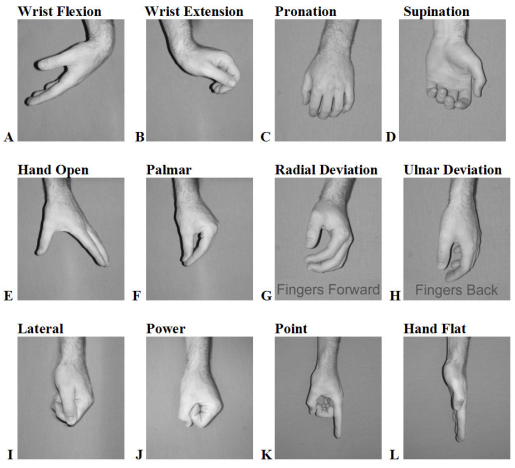

Figure 2 – The twelve movement classes investigated in this study.

RESULTS: The average classification accuracies for each of the four electrode conditions classifying all twelve movement classes are shown in Fig. 3. When only EMG amplitude was used (Fig. 3A), the targeted surface (TS), targeted intramuscular (TI) and untargeted surface (US) conditions produced similar classification accuracies ($p = 1.000$) while the accuracies resulting from the untargeted intramuscular (UI) electrodes were significantly lower ($p < 0.02$). However, the ANOVA analyses found *no* difference between the electrode conditions when the additional signal features were employed ($p > 0.05$) (Fig. 3B).

The classification accuracies for the 'clinically relevant' subsets of output classes using all available signal features are shown in Fig. 4. The repeated-measures ANOVA results showed differences between the numbers of output classes ($p<0.0005$). However, post-hoc tests from the ANOVA found no differences between the electrode conditions ($p > 0.08$ for all comparisons).

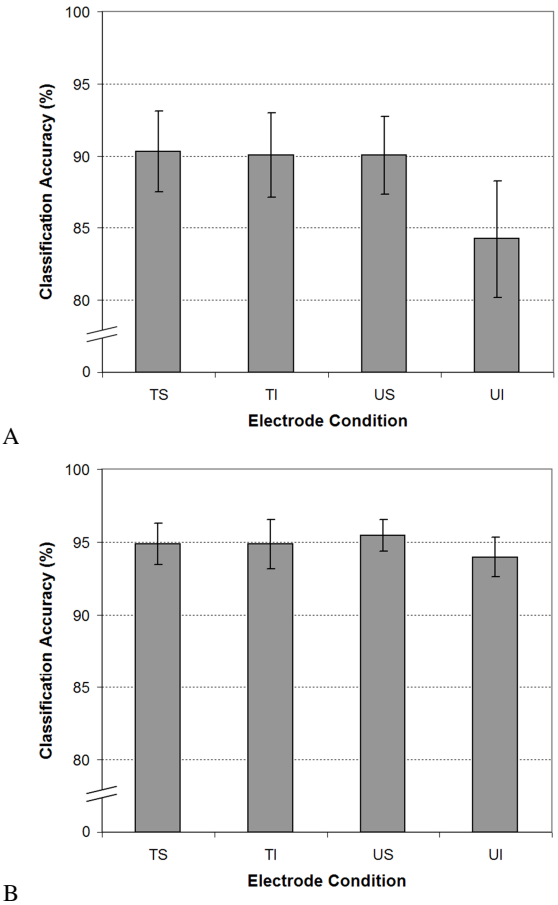

Figure 3 – Classification accuracies for the four electrode conditions for feature sets using (A) only EMG amplitude or (B) auto-regressive and time domain features as well.

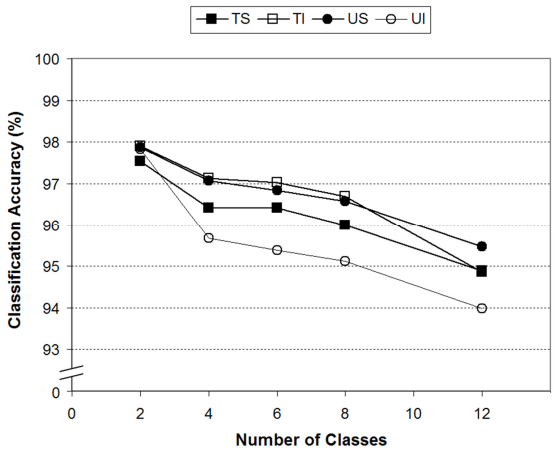

Figure 4 – Classification accuracies for different subsets of the twelve classes.

DISCUSSION: Since classification accuracies can be substantially increased for all conditions by extracting additional signal features (compare figure 3A and 3B) and the relative cost of doing so is minimal, the results from the data using all available features will be used to address the primary goals of this paper. When additional features are extracted from the EMG signals, there were no statistical differences between the electrode conditions, i.e. similar classification accuracies were obtained from both 1) surface and intramuscular electrodes and 2) targeted and untargeted electrodes.

Targeting either type of electrode did not increase the classification accuracy. This data indicates that surface electrodes do not need to be placed over specific muscles and thus socket fabrication can be simplified by arranging the electrodes in a symmetric array around the circumference of the forearm. Also, since targeting the intramuscular electrodes to specific muscles did not improve

classification accuracy, procedures such as ultrasound guidance may not be necessary to target the electrodes if a pattern recognition-based control scheme is implemented.

The similarity in the classification accuracies between surface and intramuscular recordings shows that intramuscular electrodes can perform as well as surface electrodes. Therefore, the choice of using implanted versus surface electrodes should be made, not based upon classification accuracy, but on other clinical factors. These clinical factors include comfort, cost, consistent electrode contact (i.e., surface electrode 'liftoff'), invasiveness, signal consistency with donning and doffing, signal robustness and skin impedance changes.

Class Subsets: As expected, classifiers attempting to differentiate between smaller numbers of output classes tended to have higher classification accuracies for each electrode condition. Twelve output classes were used to make the classification problem difficult in an attempt to tease out differences between the electrode conditions. However, the analyses performed on the class subsets chosen using clinical criteria showed similar accuracies between the electrode conditions regardless of the number of output classes. Therefore, the conclusions from the previous section also hold for prostheses that utilize commercially available componentry.

Summary: From the point of view of strictly maximizing the classification accuracy, the untargeted surface electrodes provided similar classification accuracies to the other three conditions and did so with the lowest cost, invasiveness and difficulty in socket fabrication. However, good clinical performance encompasses more than having a high accuracy in a laboratory setting. These experiments did not account for: variable electrode positions as the user dons and doffs the prosthesis, motion artifacts, surface electrodes 'lifting off,' skin impedance changes throughout the day, other motions of the arm being conducted (i.e., flexion/extension of the elbow) while the hand/wrist are being controlled, etc. Intramuscular electrodes have the potential to address each of these clinical issues. These other sources of noise, etc., may affect the relative performance of each electrode condition differently, which could affect the relative advantages/disadvantages of each electrode. However, this study successfully provided a comparison of the electrodes in a laboratory setting and concluded that further investigation into the clinical advantages of the different electrodes is warranted.

ACKNOWLEDGEMENTS: The authors would like to acknowledge Bolu Ajiboye, Ph.D., Ross Bogey, D.O. and Todd Kuiken, M.D./Ph.D. for their assistance as well as recognize the Coleman Foundation for the use of the Coleman Foundation Computing Center.

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EXPERIMENTAL LEAD ZIRCONATE TITANATE (PZT) SLIP SENSOR

by
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ABSTRACT

Future advanced artificial hands will require the automatic holding of objects using feedback control. To achieve this aim will require sensors of various types, one of which should be capable of detecting the relative movement between the surface of a grasped object and the hand (slip). A low-cost sensor, using thick-film technology, has been developed which detects slip using the piezoelectric effect. Experimental evaluation of the sensor has been carried out using a test apparatus whereby a block of aluminium representing an object slides past the sensor. Attached to the object surface is a Perspex sheet with repeating groves cut into the surface. Two different separations of the groves have been tested. The results show that the slip sensor detects the relative velocity between a moving object and the sensor surface. The sensor has a frequency response into the kilohertz which makes it an excellent candidate for a slip sensor. The sensor is able to detect slip with and without a cosmetic material covering the sensor. Computer simulations of the mechanical modes of vibration have shown that the frequency of the lowest fundamental mode is much higher than the electronic signal output from the sensor.

INTRODUCTION

An obvious way of measuring slip is to have a disc mounted in a hand which is in contact with the object so that when slip occurs, the disc rotates and a digital encoder produces a signal. Such a direct contact system has been used in robotics but is unsuitable in prosthetics as there is a cosmesis protecting the hand, requiring an indirect method to be found. By its very nature, a piezoelectric sensor will make an estimate of the object slip and not output a highly accurate measure. The slip signal is then either sent to an electronic controller or fed back to the wearer of a prosthesis.

FINGERTIP

Fingertip shapes were cut from 2mm stainless steel plate (type 430) to form a prototype (figure 1) [1-4]. Figure 1(a) shows a drawing of the finger tip consisting of three thick-

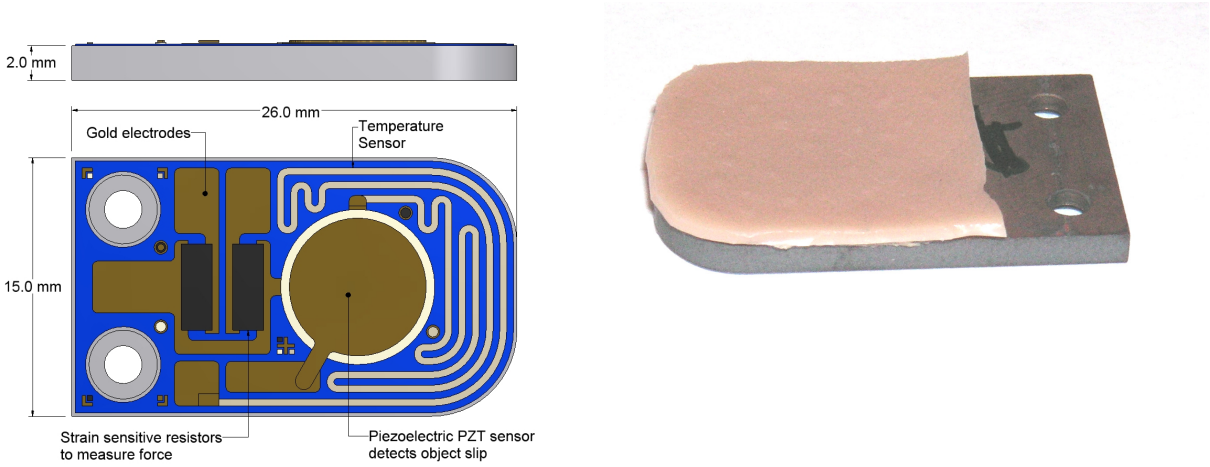

Figure 1 (a) Drawing of the finger tip sensor using a stainless steel type 430 substrate
(b) Photograph showing the back of the sensor with cosmesis covering

film sensors to measure slip, force and temperature. Towards the right of centre in the drawing is the circular piezoelectric PZT slip sensor. On the left of centre are two force sensitive resistors and arranged around the slip sensor is a temperature sensitive resistor. The top section of an Otto Bock Sensor Hand™ speed cosmetic glove was cut to the shape of the sensor and adhered to the back of the fingertip (the side without the sensors printed on) with an epoxy resin (figure 1(b)). This section of glove was selected as it is approximately 1mm thick with a uniform surface texture.

EXPERIMENTAL PROCEDURE

A sliding block mechanism was used to test the sensor. It consists of an aluminium block moving horizontally along a right-angled base plate (figure 2). The finger tip can be seen in the middle of the diagram where it is passed by a grating attached to the block. The distance moved by the block was measured using a digital rotary encoder. To accelerate the block, weights were hung from a cable moving over a pulley at the end of the base.

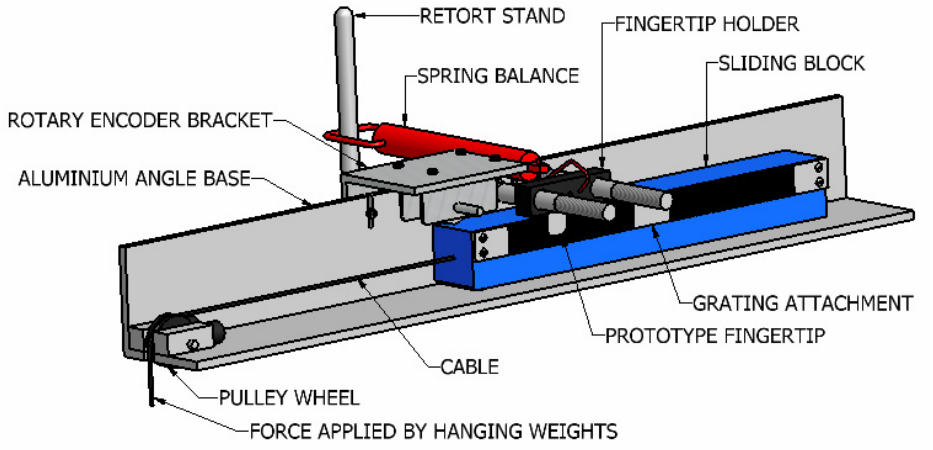

Figure 2 Diagram of the experimental apparatus

A 310mm long Perspex block attachment with two sets of grooves cut 0.1mm wide and 0.19mm deep was bolted onto the sliding block. One set of grooves, covering 109mm of the block, were spaced a distance of 1mm apart. The second set of grooves, covering the last 109mm of the block, were spaced at 0.5mm apart. To replicate slip, weights were added to the end of the wire cable in 50g increments until stiction was overcome and the block began to slide. Eight slip signals from each grating were recorded using a DAQ card and a purpose written Labview™ program. The data was subsequently analysed to estimate distance travelled by the sliding block in the first 200ms from the sensor signal and compared with the actual distance from the encoder. The average acceleration of the block was calculated and the frequency content of the slip signal determined using the power spectrum FFT function in Labview™.

RESULTS

Figure 3 shows a typical recording of a slip waveform. To estimate the slip distance from this signal it was rectified and compared to a constant threshold value. Every time the signal increased in value and crossed the threshold a count of one was recorded. On decreasing below the threshold a further count of one was recorded. These counts were accumulated and compared to the distance measured by the digital encoder (figure 4). There is a good linear

relationship between the counts and encoder data. This result is evidence that the sensor is estimating slip velocity.

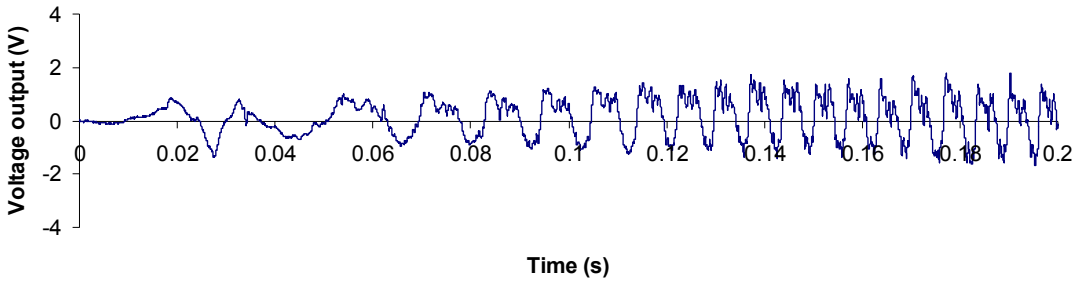

Figure 3 A typical signal from the slip sensor

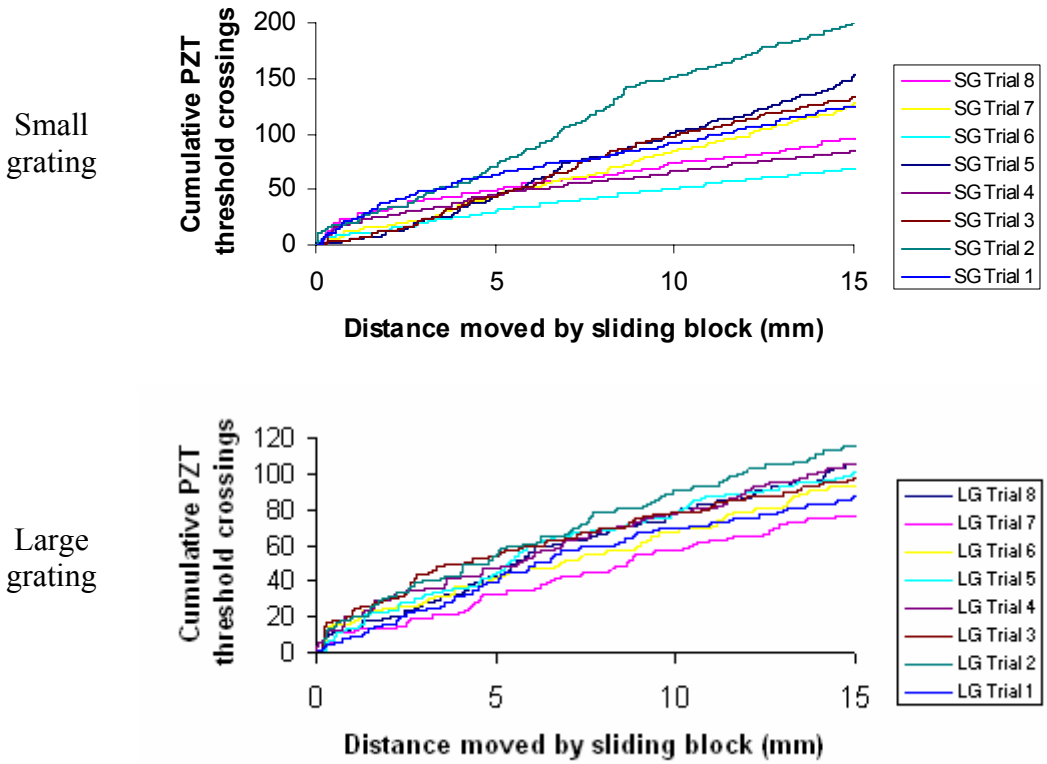

Figure 4 Estimated distance (cumulative PZT threshold crossings) against actual distance moved by the sliding block determined from the digital encoder.

Figure 5 shows the power spectrum of two trials with the two gratings where there is a similar range of frequencies. There was a wide variation in the peaks from trial to trial. There are components at slightly higher frequencies for the small grating (300 Hz) compared to the large grating (250Hz).

Figure 6 illustrates three modes of vibration for the prototype fingertip simulated using Autodesk Inventor 10™. The colour scale indicates the stress distribution caused by the fingertips vibration but does not take into account the stress caused by the deformation of the fingertip. The results from the simulations show that the lowest fundamental frequency would come from the mode shown in the left of the figure, predicting a frequency of 2kHz. This frequency is higher than the frequencies observed in the signals produced from all of the trials using both gratings. This result would suggest that the fingertip is not being excited into a fundamental mode of vibration by the grating.

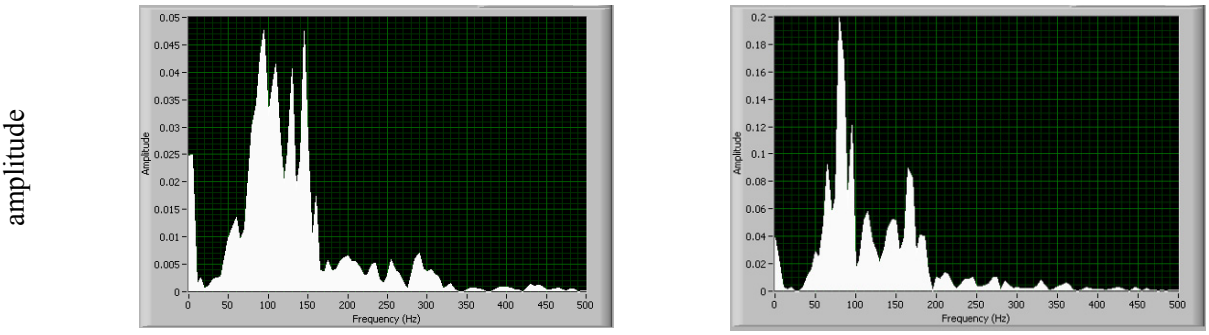

Figure 5 Frequency response of the small grating (left) and large grating (right)

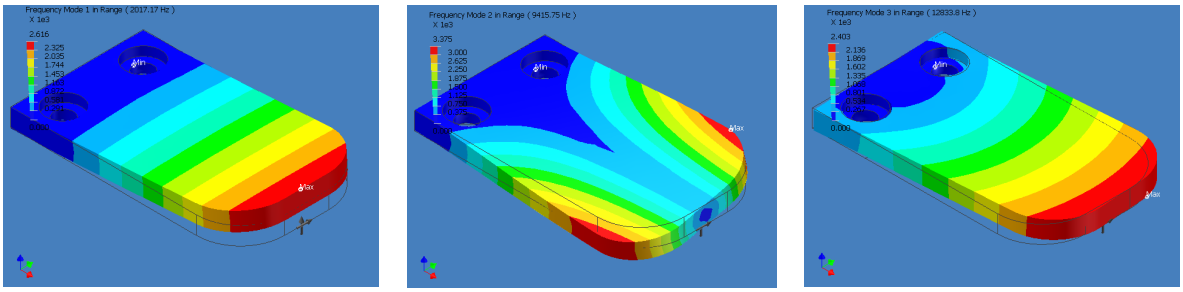

Figure 6 A simulation of the fundamental frequency modes of vibration with different constraints

CONCLUSIONS

A slip sensor made from PZT has the necessary bandwidth to estimate the distance slipped by an object when processed into a digital signal. The device operates when a piece of silicone glove, 1mm thick separates the sensor surface from an object. The frequencies of the mechanical resonance modes of vibration for the sensor are above the frequency response required for slip detection.

ACKNOWLEDGEMENT

The authors wish to thank the Engineering and Physical Sciences Research Council (EPSRC) of Great Britain for their financial support under grant number GR/R95470. Thanks are also due to the Royal Academy of Engineering for the award of a travel grant.

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ISBN: 978-1-55131-124-1