MEC17

A SENSE OF WHAT'S TO COME

Myoelectric Controls and Upper Limb Prosthetics Symposium

August 15-18, 2017

Fredericton, New Brunswick, Canada

Conference Proceedings

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Welcome to MEC17

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 MEC17. We are pleased to present a diverse and thought-provoking assortment of scientific papers and discussions relating to the field of myoelectric control and upper limb prostheses. Our theme for this year's symposium is "A Sense of What's to Come" which, aside from its direct meaning, also acknowledges the growing number of papers on sensory feedback.

Speaking of feedback, we have acted on the feedback we received during MEC14 and hope that we will continue to provide that delicate balance between a rigorous scientific program and a reunion of our tight-knit "family". In both regards, we eagerly welcome new faces to MEC.

This year's keynote speakers will highlight many of the advances in research and technology in issues pertaining to upper limb prosthetics.

- Dario Farina, PhD, is Full Professor and Chair in Neurorehabilitation Engineering at the Department of Bioengineering of the Imperial College London, UK. His research focuses on neurorehabilitation technology, neural control of movement, and biomedical signal processing and modelling.

- Jeff Tiessen is an amputee of nearly 40 years and is President and Publisher of Disability Today Publishing Group. He is a strong advocate for amputees living with limb loss.

- Doug Weber, PhD, is Program Manager in the Biological Technologies Office at the Defence Advanced Research Projects Agency, where he is managing neuroscience and neurotechnology programs.

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

We hope you will join us for the conference social events on Tuesday and Thursday, August 15th and 17th. 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 Maritime hospitality.

Once again, welcome to MEC17. Please don't hesitate to ask questions to any of our staff members.

Wendy Hill   Jon Sensinger

Co-Chairs MEC17

MEC17 Organizing Committee

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Poster Sessions

There will be three poster sessions: Session A on Tuesday, August 15th at 2:45PM; Session B on Wednesday, August 16th at 2:45PM; and Session C Thursday, August 17th at 2:45PM.

These sessions will occur during the afternoon breaks on all three days. The poster session will begin with each presenter having one minute at the podium to describe their work. Presenters will then proceed to their posters, where they will be available to answer questions until 3:45PM.

The list of posters and presenters can be found in the following program pages.

Student Paper Competition

All students who have indicated that they wish to take part in the student paper competition will have their work judged to determine the quality of their presentation and the contribution of their work to the field. Judges have been chosen to ensure fair assessment of the technical and clinical relevance of each student's work. The top three podium and poster presenters will be awarded cash prizes. Prizes will be awarded during the Banquet Dinner on Thursday evening.

Social Events

Welcome Wine and Cheese Reception
Tuesday, August 15th, 7:00PM-9:00PM
Beaverbrook Art Gallery

touch bionics BY ÖSSUR

On Tuesday, August 15th, we will host a welcoming wine and cheese reception at the Beaverbrook Art Gallery, sponsored by Touch Bionics by Össur. Self-guided tours of the gallery will be offered along with a variety of hors d'oeuvres and beverages.

Two complimentary drink tickets are provided in your lanyard.

Transportation to and from the gallery will be provided.

Banquet Dinner – Maritime Kitchen Party
Thursday, August 17th, 6:30PM-11:00PM
UNB Currie Centre

ottobock.

Sponsored by Ottobock, our popular banquet dinner will be held on Thursday, August 17th at the UNB Currie Centre.

Frantically Atlantic, a Maritime duo, will be performing during the lobster dinner, and will be inviting all to participate and learn some new instruments. They will then facilitate an open mic circle where MEC participants with "musical skills" step up to the mic. All are welcome to join in, whether to lead a song or two, or just as moral support.

A cash bar will be available. Transportation to and from the banquet dinner will be provided.

Keynote Speakers and Invited Guest Lecturer

Doug Weber
Tuesday, August 15th
9:15AM-10:15AM

Doug Weber, PhD, is an Associate Professor in the Department of Bioengineering at the University of Pittsburgh. Dr. Weber is also a Program Manager in the Biological Technologies Office (BTO) at the Defense Advanced Research Projects Agency (DARPA), where he is managing a portfolio of neuroscience and neurotechnology programs.

Invited Lecturer: Munjed Al Muderis
Tuesday, August 15th
3:45PM–4:30PM

Associate Professor Munjed Al Muderis is an orthopaedic surgeon and a clinical lecturer at Macquarie University and the Australian School Of Advanced Medicine. He specialises in hip, knee, trauma and osseointegration surgery. He is a fellow of the Royal Australasian College of Surgeons and Chairman of the Osseointegration Group of Australia.

Jeff Tiessen
Wednesday, August 16th
9:15AM-10:15AM

Jeff Tiessen, an amputee of nearly 40 years, is president and publisher of Disability Today Publishing Group (DTPG), and has been a disability community pioneer and leader for over 25 years. Tiessen is a three-time Paralympic medalist and world record holder, award-winning journalist, and Canadian Disability Hall of Fame inductee. He is a respected advocate and is keenly aware of the informational needs of Canadians with limb loss through personal experience and a vast network of amputee colleagues, O&P practitioners, and industry partners.

Dario Farina
Thursday, August 17th
9:15AM-10:15AM

Dario Farina, PhD, is Full Professor and Chair in Neurorehabilitation Engineering at the Department of Bioengineering of the Imperial College London, UK. He has previously been Full Professor and Director of the Neural Engineering Research at Aalborg University, Aalborg, Denmark (until 2010), and Full Professor and Founding Director of the Institute of Neurorehabilitation Systems at the University Medical Center Göttingen, Georg-August University, Germany (2010-2016). His research focuses on neurorehabilitation technology, neural control of movement, and biomedical signal processing and modelling. He was the President of the International Society of Electrophysiology and Kinesiology (ISEK) from 2012-2014 and is the current Editor-in-Chief of the Journal of Electromyography and Kinesiology.

Notice Regarding Audio/Visual Recording and Photography of Events

University of New Brunswick Institute of Biomedical Engineering (UNB IBME) may elect to take photographs of people and events during the MEC17 Workshops, Symposium, and Networking Events from August 15th to 18th, 2017. By attending MEC17, you agree to permit UNB IBME to use your likeness in these photos in promotion of the conference. The release checked off when registering indicated that you agree that UNB IBME shall be the copyright owner of the photographs and may use and publish these photographs. UNB IBME is released from any and all claims and causes of action that you may have now or in the future based upon or in connection with photographs and UNB IBME's use of the photographs in any manner. All rights granted to UNB IBME by you in the Release are irrevocable and perpetual. You waive all rights to any equitable relief in connection with the Release and the subject matter of the Release.

ABC Education Credits

Fifteen continuing education credits from the American Board for Certification in Orthotics, Prosthetics, and Pedorthics will be available to those attending MEC17 from August 15th to 18th. For each morning and afternoon session, a sign-up sheet will be at the Registration Desk. A Certificate of Attendance from IBME will be mailed to delegates in the fall.

In Memoriam: Professor Robert N. Scott (1933-2014)

In the time since MEC14, we lost a remarkable person. Our founder, colleague and friend, Bob Scott, passed away on December 22^{nd}, 2014. The impact of his life's work cannot be overstated; his tremendous vision and dedication created an Institute and a legacy that has enriched many lives, including the amputees that have benefited from his pioneering work and the students, guided by his extraordinary mentorship, who have become leaders themselves. Bob's passing has left a void in many of our lives, professionally and personally.

A ceremony was held on October 5^{th}, 2015, with a large contingent of Bob's family, to dedicate a tree in Bob's memory. The Bob and Joan Scott Scholarship has been established, which will be awarded annually to a deserving student entering graduate studies at IBME. Bob's beloved wife, Joan, passed away in October 2016. She was as dedicated to our institute as was Bob. The Scott family is very proud of Bob and Joan's dedication to UNB, and remain closely engaged with our Institute as his work continues to be carried out.

Professor Scott was instrumental in creating what is now MEC. In 1970, the first "Short Course" was held that would eventually evolve into MEC. It was entitled "Non-Diagnostic Electromyography in Physical and Occupational Therapy: Kinesiology and Myoelectric Control." In 1972, the first Myoelectric Controls workshop was held, with participation by orthopaedic surgeons, prosthetists, therapists and engineers. In 1993, the meeting adopted the current format of combined course and symposium.

Please join us in a moment of reflection of the tremendous contributions of Bob Scott.

Sincerely,

Kevin Englehart, PhD
Director, Institute of Biomedical Engineering
University of New Brunswick

Financial Support

The Institute of Biomedical Engineering and the MEC17 Organizing Committee gratefully recognize the following organizations for their contributions to the symposium.

TIMES	TUESDAY AUGUST 15	WEDNESDAY AUGUST 16
8:00 am	**Buffet Breakfast (60mins)**	**Buffet Breakfast (60mins)**
8:30 am	**Vendor Workshop: Ottobock (30mins)**	**Vendor Workshop: Coapt (30mins)**
9:00 am	**Welcome Address**	**Housekeeping**
9:15 am	**KEYNOTE: Doug Weber (60mins)**	**KEYNOTE: Jeff Tiessen (60mins)**
10:15 am	**Nutrition Break/ Vendor Displays (30mins)**	**Nutrition Break/ Vendor Displays (30mins)**
10:45 am	Paper Session #1 7 papers **PAIN: 3 / SENSORY FEEDBACK: 4**	Paper Session #3 6 papers **MYO CONTROL: 4 / CLINICAL PROSTHETICS: 2**
12:00 noon	**Lunch Break/ Vendor Displays (60mins)**	**Lunch Break/ Vendor Displays (60mins)**
1:00 pm	Paper Session #2 9 papers **MYO CONTROL: 4 / CLINICAL PROSTHETICS: 2 / OCCUPATIONAL THERAPY & OUTCOME MEASURES: 3**	Paper Session #4 8 papers **MYO CONTROL: 3 / PROSTHETIC DEVICES: 5**
2:45 pm	Fast-track Poster Presenters at Podium **Nutrition Break/ Vendor Displays** **POSTER SESSION A (60mins)**	Fast-track Poster Presenters at Podium **Nutrition Break/ Vendor Displays** **POSTER SESSION B (60mins)**
3:45 pm 4:00 pm 4:30 pm	**INVITED LECTURE: Munjed Al Muderis**	Paper Session #5 6 papers **OCCUPATIONAL THERAPY & OUTCOME MEASURES: 6**
5:00 pm	**End of Day Comments / RN Scott Tribute**	**End of Day Comments**
7:00-9:00 pm	**Wine & Cheese Reception** at Beaverbrook Art Gallery (*transportation provided*)	

TIMES	THURSDAY AUGUST 17	FRIDAY AUGUST 18
8:00 am	**Buffet Breakfast (60mins)**	**Buffet Breakfast (60mins)**
8:30 am	**Vendor Workshop: Touch Bionics (30mins)**	**Vendor Workshop: Infinite Biomedical (30mins)**
9:00 am	**Housekeeping**	**Housekeeping**
9:15 am	**KEYNOTE: Dario Farina (60mins)**	Paper Session #9 4 papers **MYO CONTROL: 4**
10:15 am	**Nutrition Break/ Vendor Displays (30mins)**	**10:00 a.m. PANEL DISCUSSION: 3D Printing (60mins)**
10:45 am	Paper Session #6 6 papers **OTHER: 3 / PROSTHETIC DEVICES: 3**	**Nutrition Break/ Vendor Displays (30mins)** Paper Session #10 6 papers **CLINICAL PROSTHETICS: 2 / PROSTHETIC DEVICES: 4**
12:00 noon	**Lunch Break/ Vendor Displays (60mins)**	**CLOSING REMARKS**
1:00 pm	Paper Session #7 9 papers **MYO CONTROL: 4 / SENSORY FEEDBACK: 5**	**Boxed Lunch/ Vendor tear down**
2:45 pm	Fast-track Poster Presenters at Podium **Nutrition Break/ Vendor Displays** **POSTER SESSION C (60mins)**	
3:45 pm	Paper Session #8 6 papers **OCCUPATIONAL THERAPY & OUTCOME MEASURES: 2 / CLINICAL PROSTHETICS: 4**	
4:00 pm		
4:30 pm		
5:00 pm	**End of Day Comments**	
6:30-11:00 pm	**Banquet Dinner** at Currie Center, UNB campus (*transportation provided*). Student paper awards	

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FACTORS INFLUENCING PROSTHESIS USE IN MAJOR UPPER LIMB AMPUTEES

Kristin Østlie

Innlandet Hospital Trust

ABSTRACT

Arm prostheses are an important aid to function for upper limb amputees (ULAs), and most major ULAs are fitted with prostheses after amputation. Nevertheless, the reported percentage of long-term use and the extent of actual prosthesis use in everyday life among prosthetic wearers vary considerably. Therefore, exploring and understanding factors determining prosthesis use is important to facilitate optimal prosthesis rehabilitation after upper limb loss.

We performed a cross-sectional study analyzing population-based questionnaire data (n=224) and data from interviews and clinical testing in a referred/convenience sample of prosthesis-wearing major ULAs (n=50). Effects were analyzed using linear regression, logistic regression and Cox regression.

Primary prosthesis rejection was found in 4,5%, whereas 13,4% had discontinued prosthesis use. The main reasons for primary nonuse were a perceived lack of need and discrepancies between perceived need and the prostheses available. The main reasons for secondary prosthesis rejection were dissatisfaction with prosthetic comfort, function and control. Primary prosthesis rejection was more likely in ULAs amputated at high age and in ULAs with proximal amputations, whereas secondary prosthesis rejection was more likely in proximal ULAs and in women.

Despite demonstrating good prosthetic skills, prosthesis-wearing ULAs reported actual prosthesis use in only about half of the ADL tasks performed in everyday life. Increased actual use was associated with sufficient prosthetic training and with the use of myoelectric vs. cosmetic prostheses, also in proximal amputees.

Our findings suggest that emphasizing individual needs both in prosthetic fitting and in prosthetic training is likely to facilitate successful long-term prosthesis use. These findings are incorporated in the Norwegian national guideline for rehabilitation after acquired upper limb loss, which includes strong recommendations for individualized prosthetic fitting, mandatory individualized prosthetic training and routine follow-up for prosthetic users. Also, our findings suggest that improved prosthesis quality and fitting of myoelectric rather than passive prostheses may increase long-term prosthesis use and actual prosthesis use in ADL.

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NEW EVIDENCE-BASED RECOMMENDATIONS FOR PHARMACOLOGIC TREATMENT OF PHANTOM PAIN

Kristin Østlie

Innlandet Hospital Trust

ABSTRACT

Norway's first guideline for rehabilitation after acquired upper-limb loss was published online 30 March 2016, in the open-access electronic guideline platform MAGICapp. The guideline covers all aspects of post-amputation rehabilitation, including the treatment of phantom pain. The authors were a multidisciplinary work-group, led by MD PhD Kristin Østlie. Making the guideline, we followed the steps in the AGREE-II tool. Also, we used GRADE for systematic assessment of the quality of the evidence, and for grading the strength of the recommendations (weak or strong).

The guideline has 24 recommendations for pharmacologic treatment of phantom pain after the acute postoperative phase. Several of the recommendations are weak, partly due to indirect evidence (e.g. studies on other types of neuropathic pain). This however means that the recommendations are valid for treatment of phantom pain in both upper- and lower limb amputees. The recommendations to some extent follow existing national and international guidelines for the treatment of neuropathic pain, but for some drugs, the evidence specific to phantom pain necessitated adjustment of these recommendations. Our main recommendations for phantom pain are summarized below.

We suggest that paracetamol and NSAIDs are tried before other pharmacologic agents. We then suggest the SNRI duloxetine before gabapentin, and pregabalin if these medications do not give sufficient pain relief. The evidence for the effect of duloxetine on phantom pain is of the same quality as for gabapentin and pregabalin. Duloxetine however has the advantage of not needing a time-consuming gradually dose increase, and also, the evidence suggests that the side effects for the most common (60 mg) dose are on the placebo level. Furthermore, in Norway, gabapentin must have been tried to get financial reimbursement for the use of pregabalin. The next pharmacologic agents suggested are topical lidocaine and topical capsaicin.

For peroral morphine, transdermal buprenorphine, the TCA amitriptyline and tramadol, the evidence is less convincing for phantom pain than for other neuropathic pain, and side effects are common. These are therefore not suggested as first line drugs.

For IV morphine, botulinum toxin injections, SSRI, beta blockers, memantine, ketamine, muscle relaxants (e.g. Baclofen), calcitonin, tapentadol, transdermal fentanyl, fentanyl nasal spray and local injections of corticosteroids, sufficient evidence for effect on phantom pain is lacking, and we therefore recommend against the use of these agents for phantom pain.

REFERENCE

Evidence-based guideline for rehabilitation after acquired upper limb loss in Norway [Norwegian] Østlie K (editor) et al. MAGICapp 30.03.16. www.magicapp.org/public/guideline/Jn3zaL

CASE STUDY: EXPERIENCE FITTING HEAVY DUTY STAINLESS STEEL 3D LASER SINTERED LOCKING FINGER ON A PARTIAL HAND AMPUTEE

[1]Branden Petersen C P L P, [2]Richard Weir Ph D, [3]Levin Sliker Ph D, [3]Jacob Segil Ph D and [3]Stephen Huddle M S

[1]Hanger Clinic
[2]University of Colorado
[3]Point Designs LLC

ABSTRACT

Partial hand and finger amputations make up the largest upper limb patient population when you look at upper limb amputation as a whole. In the past several years there has been many advances towards options for patients with partial hand amputations. Many of these advances can assist improving the functionality of partial hand amputees. Many partial hand patients are involved in many different activities of daily living (ADLs) that are heavy duty. In the past many of the partial hand prosthetic options do not hold up to heavy duty forces, wet,and dirty environments. Point Designs LLC recently developed a 3D laser Sintered stainless steel prosthetic locking finger designed for heavy duty ADLs. This case presentation is about a 34 year old Caucasian male that works in the building and construction industry. He Sustained a work related injury that resulted in a partial hand amputation of fingers 2-5 and thumb tip. Due to the heavy duty nature of his work a heavy duty prosthesis was required. This presentation discuses the features of this new 3D printed stainless steel locking finger and clinical application.

THE TITAN FINGER: A HEAVY DUTY TITANIUM FINGER OPTION FOR PARTIAL HAND PATIENTS

Matthew Mikosz

Hanger Clinic

ABSTRACT

This presentation will describe a new advancement in partial Hand technology. The design is a ratchet style heavy duty titanium finger called the TITAN finger that is modular and can be suitable for someone missing a full finger, partial finger or full thumb. The design operates by the user applying pressure to the dorsal side of the finger to allow the finger to ratchet into flexion. A steel pin engages into a gear that locks the finger in place. To extend the finger the user would dislocate the joint and position into a new extended position. There have been 70 devices fit in the US currently with many users who are utilizing the device for heavy duty activities and the feedback has been very positive to date. The TITAN finger has been in development for over a year and was released in mid 2016. Functional prototypes were designed and tested using a Form 2 SLA machine. The preferred method for fitting a TITAN full finger, TITAN Partial or TITAN Thumb would be to design a custom molded silicone interface and pre preg carbon fiber frame. This design offers many benefits which includes improved comfort, range of motion and durability compared to other materials. This design also allows for a very streamline socket with added strength and stability for heavy duty activities. Over the years there has been significant advancements in Partial Hand technology and the TITAN Finger can offer an additional option for someone seeking a device for heavy duty activities.

IMPLEMENTING RAPID PROTOTYPING WITH CURRENT TECHNOLOGY TO ENHANCE OVERALL FUNCTION FOR BLIND BILATERAL PATIENT

Matthew Mikosz

Hanger Clinic

ABSTRACT

This presentation will discuss the design and fitting process for a blind, bilateral partial hand/ trans-radial amputee who also underwent a face and jaw transplant after an attack by a chimpanzee and the rationale behind choosing the specific design that was implemented. After a thorough evaluation with the patient by the physician, upper limb specialist and Occupational Therapist it was determined that the patient could benefit from being fit with some of the latest myoelectric technology with some modifications to maximize her function. The technology that was chosen for her on the left trans-radial side was a myoelectric prosthesis with I-Limb Quantum, electric wrist rotator and custom silicone socket interface. The right partial hand was a custom made implement holder that was designed in CAD and 3D printed. The purpose of this presentation is to highlight the benefits of rapid prototyping and how implementing with current technology can enhance the functional outcomes for the user. Utilizing the I-Limb Quantum with electric wrist rotation had proven to have many benefits that improved her activities of daily living but also presented some challenges. The wrist rotator presented the most challenges as being blind she was unable to determine where the hand was in space. This challenge was anticipated and led to the design of a wrist rotation limiting device that was placed in between the hand and prosthesis to limit the motion of the wrist rotator. The range of motion determined to be required for functional activities was around 90 degrees. This allowed her to always know where the hand positioning was based on an audible beep from the wrist rotator when the limits were achieved. The right partial hand device was designed specifically for her to better assist her with feeding herself, writing and grooming. Several designs were developed over the week and incorporated into her therapy sessions. At the end of the week the optimal design was printed and implemented into her prosthesis. This fitting has shown that currently available technology alone can have great benefits for the user but when limitations or challenges arise some modifications or additions can compound the benefits and overall success for the user.

Utilizing Wrist Movement for Prehension in an Atypical Case of Bilateral Partial Hand Amputation

Phil Stevens MEd, CPO

Hanger Clinic, USA

ABSTRACT

This case study presents upon the unique utilization of wrist movement for prehension in an atypical case of bilateral partial hand amputation. The patient in question presented with a single residual metacarpal on both extremities. These digits were sensate. Mobilization was limited to wrist flexion with limited extension range. Recognizing the value of these sensate digits and the remaining active motion of the distal extremities, a two-position APRL thumb was placed on the ventral aspect of the wrist to provide an opposition post to the residual digit. This approach was used bilaterally, permitting bilateral active prehension with a sensate digit.

This concept was further exploited in the fabrication of custom silicone restoration prostheses in which the individual digits mobilized the metacarpal region of the hands bilaterally while the silicone thumbs were stabilized against the ventral aspects of the forearms.

The prosthetic solutions enabled the patient to benefit from the residual sensation and mobility afforded by his unique presentation to produce bilateral active prehension.

INTRODUCTION

Service missionaries in Uganda, Africa identified an individual with atypical bilateral traumatic partial hand amputations. After failing to identify capable resources in his home country, the patient was brought to the United States for prosthetic rehabilitation. The patient's presentation included bilateral preservation of a single metacarpal (suspected to be the first but possibly the second). These residual digits retained the advantage of sensation and could be actively mobilized using wrist flexion (Fig 1).

Following a remote Skype-assisted evaluation, attempts were made to identify local resources that might provide acceptable prosthetic care. When these could not be identified, the decision was made to bring the patient to the US for prosthetic rehabilitation.

Figure 1: Active wrist flexion range of motion with a single distal metacarpal digit bilaterally

PRIMARY PREHENSILE DEVICES

Recognizing the value of exploiting the active mobilization provided by the wrist joints and the preserved sensation of the residual digits, the use of a two position thumb mounted proximal to the wrist joints was explored with wrist flexion closing the grip and wrist extension opening the grip (Fig 2).

Figure 2: Simulation of wrist-enabled prehension with the wrist joint extended and flexed.

With this approach validated, test sockets were fabricated to allow dynamic use of this prehensile approach (Fig 3). These were constructed of fiberglass, molded directly over flexible inner sockets with the two position APRL thumb units epoxied to the fiberglass frames.

Figure 3: Demonstration of tip prehension and writing using a test socket prosthesis.

With the patient demonstrating functional prehension bilaterally, these devices were subsequently manufactured definitively. In the definitive construction, the flexible sockets were trimmed distally to the wrist crease to augment the exposed skin and associated sensory input as well as the mobility of the residual digits. Leather sleeves were fitted to the thumb component to increase the localized coefficient of friction and subsequent prehension (Fig 4).

Figure 4: Definitive fabrication of the primary prehensile prostheses

SECONDARY COSMETIC PROSTHESES

In addition to the primary prehensile devices, the patient was concerned about the cosmetic appearance of the partial hand amputations and the associated limitations related to societal interactions and educational and vocational opportunities. Accordingly, custom silicone restorations were additionally fabricated. In doing so, attempts were made to facilitate a degree of active prehension using the principles described above.

In the evaluating the dynamic fit of inner silicone sockets, material bunching in the volar aspect of the wrist was observed. This was addressed by removing a wedge of silicone in the region to permit full flexion mobility at the wrist (Fig. 5).

Figure 5: Removal of a silicone wedge at the ventral aspect of the wrist permitted additional wrist flexion mobility in the silicone inner socket

This done, the external cosmetic silicone elements were constructed. The metacarpal region of the restoration was fabricated over the residual digits bilaterally, the finger restorations extending distally. The thumb restorations were brought proximal to the wrist such that wrist flexion reduced the distance between the silicone fingers and the silicone thumb (Fig 6). This created a degree of active opposition that allowed for limited prehensile function.

Figure 6: Demonstration of active prehension with the silicone restoration prostheses using active wrist motion, including holding a partially filled cup of water

HEAVY DUTY PROSTHESES

Recognizing that both of the prosthesis types described to this point would best be described as light to medium duty solutions, a final set of body powered prostheses with heavy duty terminal devices were also provided. These were controlled by individual figure-9 control harnesses allowing the patient to mix devices on task specific bases (Figure 7). Tamarak Flexure joints at the radial aspect of the wrists preserved wrist flexion mobility. This flexion, combined with residual pronation and supination permitted reasonable prepositioning of the body powered terminal devices

Figure 7: Patient utilizing his primary prehensile device on his left extremity and a heavy duty body powered device on his left extremity

CONCLUSION

Following prosthetic fittings, the patient returned to Uganda. As expected, the prehensile devices became his primary resource for upper limb function. He also reported use of the heavy duty prostheses for more aggressive tasks like gardening and hauling water. The silicone prostheses were rarely worn with the patient citing discomfort. Following the departure of the Uganda-based Service Missionaries, the patient was lost to further follow up.

ACKNOWLEDGEMENTS

Paul Tanner fabricated the Silicone Prostheses

SYSTEM GROUP DYNAMICS AND THEIR EFFECT ON UPPER LIMB INNOVATION IN O & P

Gerald Stark, Ph.D, MSEM, CPO/L, FAAOP

Coapt, LLC

INTRODUCTION

This study examines the effects of group dynamics and social interaction with innovation among orthotist and prosthetists. Since any new innovation inherently comes with a higher degree of uncertainty and risk, the group or individual must deal with the anxiety created by innovative behaviour. Individuals whoa re less anxious and risk adverse may tend to adopt innovations more easily than others who regard changes with greater uncertainty. Individuals can be classified into adopter categories based on their rate of adoption of new technologies and capacity of risk and anxiety. Individuals who are more susceptible to anxiety in general, may seek the emotional scaffolding of their organizational group to support innovative behavior. This may be especially true in healthcare where contextual stress is heightened.

METHOD

The intent of this study was to examine if there is any relationship between an individual's differentiation of self and level of technology readiness. The level of differentiation within the work context will be compared to innovation technology readiness. This study construct was a non-experimental, associational, design using an electronic survey comparing emotional differentiation, as measured by the Workplace Differentiation Inventory (WDI), and technology readiness as measured by the Technology Readiness Index 2.0 (TRI-2.0). The intent of the study was to examine the potential relationships between the WDI and TRI-2.0 as well as the subattributes of both instruments. The analysis was done to find if any relationships exist between with demographic attributes of gender (G), years of experience (EXP), professional certification (CERT), technology self assessment (TSA), number of high-tech patients per year (HTP), number of external linkages (EXLK), number of internal linkages (INLK), and professional affiliation (AFF).

RESULTS

The survey, which included the eight demographic questions as well as the WDI and TRI 2.0, was made available with a link and invitation on the OANDP-L list server. The survey was posted on Qualtrics from August 18, 2015 until August 31, 2015, and had n = 148 respondents. Examination of the relationships using two-tailed Person's correlations showed significance between Technology Optimism with all attributes of the WDI; Fusion with Others, Emotional Reactivity, and Emotional Cut-off. Technology Innovation also had significant relationships with Fusion with Others, Emotional Reactivity, and Emotional Cut-off. The regression analysis showed a moderately strong predictive relationship between the WDI and the TRI-2.0. A very strong predictive relationship was found between Technology Optimism with Emotional Cut-off and Emotional Reactivity. Technology Optimism and Emotional Reactivity alone shared a strong predictive relationship. Conversely, the WDI had very strong predictive relationship with Technology Optimism, Technology Innovativeness and Technology Insecurity with Technology Optimism contributing a majority of the effect. An extremely weak relationship between the WDI composite score and Years of Experience.

DISCUSSION

This study has shown that Emotional Reactivity and Emotional Cut-off had a significant predictive relationship with Technology Optimism. This study has also shown that Technology Optimism, Technology Innovativeness, and Technology Insecurity had a very strong significant predictive relationship with Workplace Differentiation, specifically Emotional Reactivity. The other key result was that Gender, Technology Self-Assessment, Certification Level, Years of Experience, and Office Affiliation had little or no effect on the measures of differentiation or technology readiness. The implication is that continual introduction of new concepts and technology would be a strong predictor of a less emotionally reactive and thoughtful group for change represented by technologic and reimbursement advancements.

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FACTOR ANALYSIS OF UPPER EXTREMITY PROSTHETIC PATIENT ACCEPTANCE

Gerald Stark, Ph.D, MSEM, CPO/L, FAAOP

Coapt, LLC

INTRODUCTION

Although there have been a number of studies that have attempted to investigate prosthetic upper limb prosthetic acceptance, very few have provided conclusive evidence to assess the overall clinical probability of wear. In a recent survey of upper limb prosthetic practitioners' collective estimations were made with respect to the priority of the various factors that contributed to Upper Limb rejection as well as the rejection rates by level. From these factors a Bayesian forecasting model was constructed to provide an initial estimation of prosthetic acceptance.

METHOD

From initial phone interviews and previous upper limb survey 12 factors were identified as possible contributors to overall acceptance including: Amputation Level, Functional Advantage, Patient Gadget Tolerance, Time to Initial Fit, Confidence of Prosthetist, Quality of Patient/Prosthetist Relationship, Socket and Harness Comfort, Weight, Cosmetic Quality, Therapy and Training, Peer and Family Support, and Age of Patient.

A survey was constructed and posted on a third-party survey hosting website from October 20 to November 8, 2013. There were 58 respondents who self-assessed their skill level which was normally distributed among novices, intermediates, experts, and specialists.

RESULTS

Their aggregate responses with respect to level were recorded to be 79.6% for transrandial, 57.8% for transhumeral, and 32.8% for shoulder disarticulation which is fairly similar to the original study done in 1958 which indicated 75% for transhumeral, 61% for transhumeral, and 35% for shoulder disarticulation [2] When asked about the degree of strength each factor affects acceptance the aggregate opinion was that "Amputation Level" was the highest at 79.6, followed by "Functional Advantage" at 78.3, "Socket and Harness Comfort" at 77.7, "Peer and family Support" at 76.3, "Amount of Therapy and Training" at 73.5 and "Quality of Patient-Prosthetist Relationship" at 72.6.

DISCUSSION

From this survey of the 12 factors those with the most consistent significance are listed in order: 1) Amputation Level, 2) Functional Advantage, 3) Socket & Harness Comfort, and 4) Peer/Family Support and 5) Prosthetic Competency. Many examinations fail to also include the important role of the peer and family support as well as the prosthetist-patient relationship. This could represent the consistency of smaller, but more innovative populations to accept technology.

CONCLUSION

There are methods of functional prediction in lower extremity prosthetics such as the AMPPRO Ambulation Test and Amputee Mobility Detector (AMP) [3] that utilize greater degrees of probability and forecasting. One factor is that clinicians and researchers have both attempted to find a single factor that can influence success or failure. The characteristic of the high rating for all of the factors may indicate that clinicians were unable to delineate between the factors and rank those with greater probability. This could be that different combinations of factors greatly vary from patient to patient, that prosthetists differ widely in their opinions about rejection, or that prosthetists in general do not have a grasp of why rejection occurs. There appears to be some discrepancies as to whether rejection and acceptance are converses of each other. The question remains why acceptance rates have been largely unchanged and vary so greatly.

CONCLUSION

The amputation acceptance levels were used with the other factors using a 4-point value scale of .10, .25, .50, or .75 and calculated using Bayes' Theorem for a number of case studies. Although not validated they may serve as analytic tool to assess subjective values of acceptance.

FIGURE AND TABLES

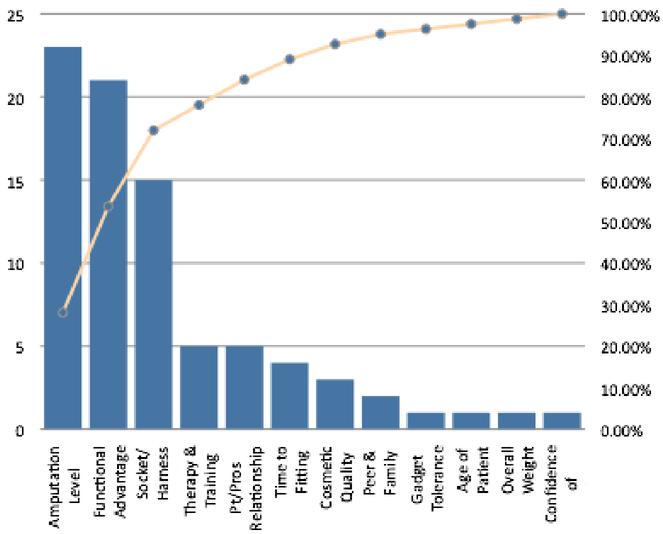

Figure 1: Pareto analysis of acceptance

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FACTORS THAT INFLUENCE ACCEPTANCE AND REJECTION OF AN UPPER LIMB PROSTHESIS: A REVIEW OF THE LITERATURE.

Andreas Kannenberg

Otto Bock Healthcare LP

INTRODUCTION

Clinicians and health insurances are well aware of the fact that many patients with upper limb (UL) amputations reject their prosthesis in the mid- to long run. Factors that influence acceptance and rejection of an UL prosthesis are much less understood. If such factors and their impact were known, they could be leveraged to improve the acceptance of UL prostheses and the function and quality of life of persons with UL amputations.

METHOD

A search of the scientific literature was performed in the Medline, Embase, CINAHL, OTseeker, and PEDro databases as well as in the online library of the Journal of Prosthetics & Orthotics. Search terms were related to UL amputations and prosthetics, acceptance, use, rejection and abandonment of UL prosthesis. Identified references were evaluated for pertinence to the subject and analyzed.

RESULTS

Malone et al. suggested a “golden window” of 30 days after the amputation for the fitting of an (interim) UL prosthesis for occupational therapy. They found that all patients who received a prosthesis within this “golden window” were able to return to work, whereas only 15% of patients fitted after more than 30 days did so. In addition, patients fitted within the “golden window” did not present any striking preference for body-powered or myoelectric prostheses, whereas patients who were fitted later almost exclusively preferred myoelectric prostheses. Another study found that definitive prosthesis fitting within 6 months of the amputation or 2 years after birth in congenital deformities increased the likelihood of prosthesis acceptance (odds ratio) by factor 16. The second biggest variable was the involvement of the patient in the selection of the type of prosthesis. Intense patient involvement increased the likelihood of acceptance by factor 8. Also, patients with transradial amputations were more likely to accept a prosthesis than patients with more distal or proximal levels of limb absence.

DISCUSSION

Patients should be fitted a prosthesis for occupational therapy as soon as medically possible, ideally within 30 days after the amputation to prevent them from learning to manage their everyday lives with their sound hand alone. Definitive prosthesis fitting should occur within 6 months of the amputation or 2 years of birth in case of congenital deformities for the same reason. Also, patients should be intensely involved in the selection of prosthesis type as they are then 8-times more likely to accept their UL prosthesis than those not involved in decision making.

PROVISION OF ACTIVE UPPER LIMB PROSTHESES AROUND THE WORLD.

Andreas Kannenberg

Otto Bock Healthcare LP

INTRODUCTION

A recent systematic review of the literature has shown that there is no evidence for a general functional superiority of body-powered or myoelectric/externally powered prostheses, but cosmesis and appearance of myoelectric hands were significantly better [1]. The purpose of this paper is to give an overview on the perspective of different health care systems around the world on the coverage of active upper-limb prosthetics.

METHOD

As no authoritative statistics are publicly available, data was collected by interviewing acknowledged professionals in the field of upper-limb prosthetic rehabilitation as well as by obtaining estimates of market sizes for body-powered and myoelectric/externally powered upper-limb prostheses from Ottobock´s business unit, national market managers, and clinical prosthetists specialized in upper-limb prosthetics in various countries.

RESULTS

Countries may be placed in one of three categories upon the basic approach to the provision of active upper-limb prostheses:

1. Health care systems that grant access to all types of active upper-limb prostheses (e.g. Western and Northern Europe). In these countries, myoelectric/externally powered upper-limb prostheses are considered standard of care. However, adoption levels vary depending on differences in coverage policies. In countries whose healthcare systems cover several upper-limb prostheses at a time, 80-90% of patients use myoelectrics as their primary prosthesis. If the healthcare system covers only one prosthesis at a time, the proportion of myoelectric prostheses may decline to 50-60%.

2. Health care systems that limit access to active upper-limb prostheses primarily to body-powered devices (e.g. USA, Canada, Australia, New Zealand, Japan). In these countries, myoelectric/externally powered prostheses require approved exceptions from coverage policies and are typically used by less than 35-40% of all patients with active prostheses. However, in some specialized urban rehabilitation clinics, the share of myoelectric prostheses may reach up to 70%.

3. Countries with health care systems that provide passive or no upper-limb prostheses to the vast majority of their beneficiaries (e.g. Eastern Europe, Latin America, Asia, Africa).

DISCUSSION

Although the scientific evidence for upper-limb prosthetics is the same around the world, coverage policies and funding vary remarkably and result in strikingly different adoption rates of active upper-limb prosthetic technologies between different countries. Among industrialized countries, the most important difference seems to be whether policies only consider prosthetic function or also psychosocial aspects for determining medical necessity of the available active prosthetic technologies and designs. Unfortunately, no data is available to conclude which strategy results in higher prosthesis acceptance rates.

A FOCUS ON THE PATIENT EXPERIENCE: ADVANCED UPPER LIMB PROSTHETIC RESTORATION VS HAND TRANSPLANTATION AND TOE-TO-HAND TRANSFERS

Diane Atkins, OTR, FISFO, Assistant Clinical Professor

Department of Physical Medicine and Rehabilitation, Baylor College of Medicine, Houston, Texas

INTRODUCTION

Dramatic advances have been made in electric multi-articulating hands, hand transplantation and reconstructive hand surgery during the last several years. When debating the best solution, it is critically important to enable individuals with limb loss to be able to make an informed decision with respect to aspects of: time from procedure to “function”, costs, amount of therapy required, medications, potential complications, sensation, pinch and grasp, functional outcomes, as well as the appearance of the hand. The purpose of this study is to present the experience of individuals with bilateral hand amputations, their perception of disability as well as their function following these interventions.

METHODS

The subject population included 3 study groups- 3 bilateral transradial users of electric multi-articulating hands, 4 bilateral hand transplant patients and 1 bilateral multiple toe-to-hand transfer patient. Each individual was evaluated with the Southampton Hand Assessment Procedure (SHAP) and the Disabilities of Arm, Shoulder and Hand (DASH).

RESULTS

The Index of Function, as defined by the Southampton Hand Assessment Procedure (SHAP), demonstrated surprisingly similar results among bilateral prosthetic users, bilateral recipients of hand transplantation, and the bilateral toe-to-hand transfer. When comparing the results of the Disabilities of Arm, Shoulder and Hand (DASH), bilateral transradial users of electric multi-articulating hands scored a lower perception of disability (mean=39.83) when compared to individuals who had undergone hand transplant surgery (mean=53.25). The DASH score of the individual who had undergone bilateral toe-to-hand transfers was the lowest at 27.

CONCLUSION

Although the subject sample is small, this study sets the stage for further investigation as advances and options become available for the individual who has lost both hands, so that a prospective patient can compare not only the objective functional outcomes, but also the subjective experience as well.

REVIEW OF THE CURRENT LITERATURE ON THE CLINICAL BENEFITS OF MULTIARTICULATING PROSTHETIC HANDS.

[1]Andrea Giovanni Cutti and [2]Andreas Kannenberg

[1]Centro Protesi INAIL
[2]Ottobock

INTRODUCTION

For more than 40 years, myoeletric prosthetic hands have only allowed for the tripod opposition grip. In the past 10 years, multiarticulating hands offering up to 36 different grips have become available and popular among patients and clinicians (1). Therefore, a review of the literature on the clinical benefits of multiarticulating hands appears warranted.

METHOD

Scientific literature was searched in the Medline, Embase, CINAHL, OTseeker, and PEDro as well as in the online library of the Journal of Prosthetics & Orthotics. Search terms were related to multiarticulating prosthetic hands and their clinical benefits. Identified references were evaluated for pertinence to the subject and then analyzed.

RESULTS

Only three publications, one case study (2) and two clinical studies (3, 4) on the clinical benefits of multiarticulating hands could be identified. The case study (2) was conducted with the iLimb in a 45-year old man with a wrist disarticulation and concluded that it had only limited additional functionality compared to the DMC plus hand (2). The two clinical studies were both conducted with the Michelangelo hand. A survey with the OPUS-UEFS for perceived function in 16 transradial amputees demonstrated improved ease of performing activities of daily living (ADL) and increased active use of the multiarticulating as compared to standard myoelectric hands (3). A study with 6 transradial amputees assessing performance-based outcomes measures found significant improvements in the SHAP, the Box and Blocks test, and the Minnesota Manual Dexterity Test (4). Patient interviews after 6 months revealed enhanced perceived functionality and the perception of Michelangelo “as a real hand”, resulting in improved integration of the prosthesis into the body image (4).

DISCUSSION

The body of published evidence for the clinical benefits of multiarticulating prosthetic hands is still very limited. Two studies have found significant improvements with the Michelangelo hand. No studies have been found with multiarticulating hands that offer even more grip patterns. Therefore, it remains unclear if the clinical benefits of all multiarticulating hands are comparable or if there is a correlation between the number of available grip patterns of a hand and the magnitude of clinical benefits it may deliver.

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DISCLOSURE

Andreas Kannenberg is a full-time employee of Otto Bock.

CUSTOM SILICONE SOCKET USER SURVEY

Jack Uellendahl and Joyce Tyler

Hanger Clinic

ABSTRACT

The use of High Consistency Rubber (HCR) silicones has been shown to be clinically advantageous for use in upper extremity prosthetics [1,2,3]. Until now, no formal review of patient feedback has been reported. Anecdotally, one of the primary reasons that these custom silicone sockets have been preferred by users is the improved comfort they afford. Prosthesis discomfort is often associated with prosthesis abandonment. In order to better understand user's impressions and to determine if wearers do indeed find custom silicone sockets to be more comfortable than non-silicone sockets, a survey was developed. This survey was administered to 25 upper-limb amputees fitted by eleven different Hanger Clinic Upper Limb Specialists.
Amputation levels including six wrist disarticulations, sixteen transradials and four transhumeral amputation were represented including one bilateral with wrist disarticulation and transradial amputations. Prosthesis types represented were three passive, seven body powered, and seventeen myoelectric (two users were provided with two types of prostheses using HCR silicone sockets). Eighteen of the twenty-five users reported the silicone socket to be more (n=7) or much more (n=11) comfortable than their previous non-silicone socket. While four reported the same comfort and one user reported less comfort, none reported much less comfort.

INTRODUCTION

Socket designs using HCR silicone technology have been previously described (figure 1) [3,4]. HCR silicone offers several different material characteristics that should improve user comfort compared to non-silicone socket materials including; elasticity, multiple stiffness options that can be seamlessly blended to achieve a customized compression profile, and high coefficient of friction for better socket retention. A survey was developed to seek user input regarding their perceptions of prosthetic sockets made using HCR silicone techniques compared to non-silicone sockets. All of the users were clients of Hanger Clinic. Inclusion criteria was wearers of upper limb prostheses who had been fit with a HCR socket and previously wore a non- silicone socket. In a review of our patient records, one hundred and sixty persons were found who met the inclusion criteria. A total of twenty-five users completed the questionnaire during the period from September 2016 through February 2017. Verbal informed consent was secured from each respondent prior to the administration of the questionnaire. All levels between wrist disarticulation and shoulder disarticulation were accepted. Survey respondents included fourteen males and eleven females, seven were congenitally limb deficient and eighteen had acquired amputations. The age range was from fourteen to seventy nine years old with a mean age of fifty three. Regarding prosthesis wear experience, we found that users reported having worn a prosthesis from between one to sixty years with a mean of twenty-three years. Silicone sockets had been worn from six months to eleven years with a mean of four years. Daily wear time was reported to be from occasional use to eighteen hours a day with the majority, seventeen users, reporting daily wear.

RESULTS

Custom silicone interfaces have been noted to provide protection for fragile skin in such cases as severe burns and in the presence of skin grafts [5]. Two questions were asked to shed light on the topic of skin irritation and pain. When asked: Since switching to a silicone socket, have you experienced {more, less, same, didn't have before} skin breakdown or irritation? Three, 13%, users reported more breakdown or irritation, while six, 27%, respondents reported less breakdown or irritation. The remainder, 60%, reported either the same breakdown or irritation or they didn't experience breakdown or irritation with their previous prosthetic socket. Significantly, all three who reported more breakdown or irritation also reported that their silicone socket was much more comfortable than their previous non- silicone socket. When asked: Since switching to a silicone socket, have you experienced {more, less, same, didn't have before} pain? Only one user, (5%), reported more pain and eight users, (36%), reported less pain. Again the remainder didn't have pain or the pain was same as with the non- silicone socket.

Prosthesis wearers often comment about the temperature of their limb being hot. There have been attempts, currently and in the past to address heat retention by changing the socket material or modifying the material to reduce heat retention. Therefore users were asked to compare their silicone socket to their non-silicone socket in this regard. Four individuals, (17%), reported that their residual limb felt cooler. Ten, (42%), felt it was warmer, and another ten said it was the same. Interestingly, eight of the users who reported the socket to be warmer also stated that the socket was more comfortable than their non-silicone socket.

USER COMMENTS

"This is the best system that I have ever used and I have used them all. I can keep it turned on and I don't have to worry about it activating when I don't want it to or falling off when I perspire."

"I sweat a lot. The silicone socket holds the sweat in. Silicone itself is hot. I dump the sweat out. I like the silicone socket. I would not go back to the old socket."

"I love love love my arm. Closest thing to having a real arm. It's part of me."

"My newer myoelectric arm is much more comfortable and lighter than my previous one without the silicone."

There was lots of migration in the skin (with non-silicone) and the silicone socket fits much better than non-silicone.

"I prefer the non-silicone socket because it's more rigid and loses electrode placement from using the silicone socket, it's too loose."

"Nicer in cold weather. Retains body temp."

"Maybe the same or a little warmer. I do not seem to sweat in my silicone socket, but it is not summer yet."

CONCLUSIONS

This survey confirms our hypothesis that HCR custom silicone upper limb sockets provide more comfort and are preferred by users (Table 1). The survey reveals that users of HCR silicone sockets tend to have less skin irritation and less pain than with their non-silicone sockets if they experienced irritation or pain prior to being fit with silicone. Although there is a definite tendency to perceive that the silicone is warmer, further investigation in that specific topic is needed to better understand this issue. Some of the user comments suggest that even though the silicone socket was perceived to be warmer, the presence of perspiration was less of a problem in the silicone socket as long as the socket was not too loose. Based on our clinical experience and the results of this user survey, we recommend that HCR custom silicone socket technology be considered for all levels of upper limb prostheses.

Table 1: Results of comfort question

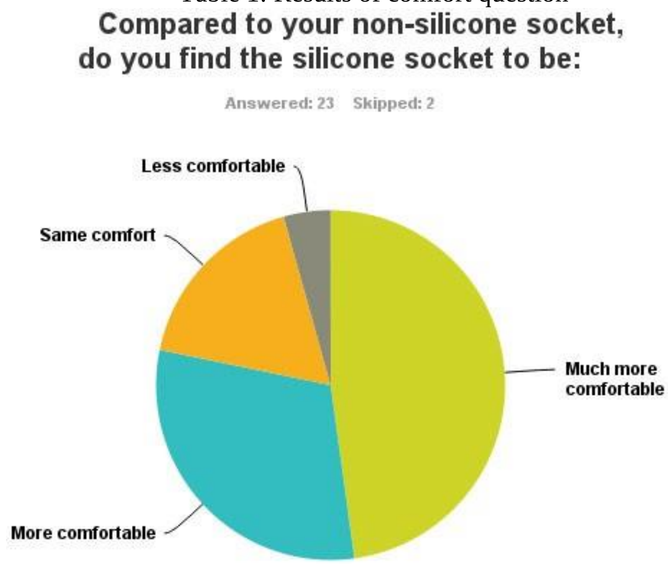

Figure 1: Example of a HCR silicone socket for transradial prosthesis with ¾ design. The brim dimension is maintained by incorporation of a carbon strut integrated between layers of silicone. The transhumeral example on the right shows the silicone socket removed from the prosthesis during trial.

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A PEDIATRIC SHOULDER DISARTICULATION/PARTIAL HAND: CASE STUDY AND SIX YEAR FOLLOW-UP

Jack Uellendahl

Hanger Clinic

ABSTRACT

Management of high level pediatric limb deficiency is challenging. Issues of prosthesis weight, complexity of control and functional utility are often cited as reasons for prosthesis rejection. Design goals include lightweight construction, simple control, and functional grasp [1]. This case study will review the prosthetic treatment over a six year period of an individual who presents with multiple congenital physical anomalies. Our subject, MK, presents with absence of his right arm at the shoulder disarticulation level, left partial hand with complete absence of his thumb, fused left elbow at sixty degrees of flexion, no forearm rotation, and scoliosis. This case presentation demonstrates that in cases where the prosthesis can provide functional gain, is light-weight and simple to control, the high level congenital limb deficient individual can achieve long-term success with appropriately designed prostheses.

INTRODUCTION

MK received his first prosthesis at age four. This prosthesis had passively positioned elbow and shoulder with a split hook. This prosthesis did not provide active grasp but did allow for passive function such as holding down objects while manipulating with the left hand or carrying objects draped over the forearm with the elbow flexed. Progression to an active terminal device was later attempted by attaching the control cable to a thigh cuff. Terminal device activation was possible by leaning or rotating the trunk however MK did not find that this control produced meaningful terminal device function. Despite the lack of dynamic grasping function, MK still wore the prosthesis daily at school.

At age seven MK was referred to Hanger Clinic seeking a more functional prosthesis. At this time it was decided to fit a myoelectrically controlled hand that MK could control with a single site control scheme using his pectoralis muscle. Reduction of prosthesis weight and the ability to position the myoelectric hand in a wide variety of locations were considered to be primary goals in order to maximize the utility of the prosthesis. Additionally, the absence of a thumb on the left hand was addressed by provision of a multiposition thumb (figure 1).

Figure 1: Shown at first test fitting of hybrid prosthesis.

The right shoulder disarticulation prosthesis employed the Otto Bock System 2000 hand with voluntary open/automatic close single site control strategy. The LTI Omni wrist provided multi-direction angulation of the terminal device passively positioned under static friction. The LTI/Steeper friction elbow allowed elbow flexion and rotation with conduit through the elbow for the control signal. The Otto Bock ball joint shoulder provided multi-direction positioning of the shoulder. The forearm, humeral section, and socket frame were made of prepreg carbon to minimize overall weight. The socket interface was made of HCR silicone. The finished weight of the completed prosthesis was 680 grams (figure 2).

The left prosthesis used a HCR custom silicone construction with carbon prepreg stabilizer for the Vincent Systems locking finger. The finger provides multiple positively locking positions and is repositioned by distracting the joint, then moving it to the desired position and releasing the finger (figure 3).

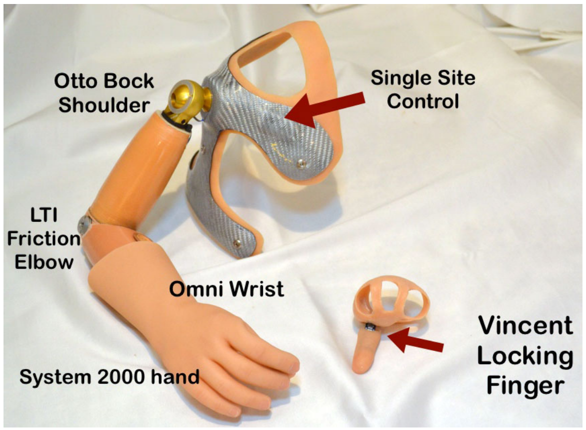

Figure 2: Components of MK's prostheses.

After receiving his new set of prostheses MK commented that the prosthesis was much lighter than his previous right prosthesis and he liked the more life-like hand compared to the hook and how small the socket was compared to his old prosthesis. He was immediately capable of controlling the myoelectric hand and was able to reposition the left thumb using the right hand to actively grasp the thumb. The multiposition thumb provided the ability to grasp objects of large or small diameter depending on the position of the thumb. At a recheck appointment two months after delivery, MK reported that his wear time had increased to a maximum of 8 hours and he continued to wear the prostheses daily at school.

One year after delivery, both prosthetic interfaces were remade due to growth reusing all of the existing components. At a recheck appointment two and a half years after the myoelectric prosthesis was delivered, MK is now ten years old and reports continued daily use of the shoulder disarticulation prosthesis. He has discontinued use of the thumb prostheses for the present time. MK feels that he can do a lot without the thumb prosthesis and sometimes it gets in the way. Since he is not able to independently donn the thumb prosthesis he feels it is easier not to wear it. He does seem interested in using the thumb prosthesis for specific activities.

Figure 3: Repositioning thumb using myoelectric hand.

Due to his fused elbow, MK is unable to reach the superior harness strap of his right prosthesis and therefore cannot independently don it. Our future plan is to explore different buckle systems such as those from Fidlock that have a magnet to allow easy one-handed attachment. MK has been able to doff the shoulder disarticulation by having a string attached to the superior strap that hangs low enough for him to reach it with his left hand. He is independent in doffing the left prosthesis by wedging it between his knees and pulling it off.

CONCLUSTION

This case presentation demonstrates that in cases where the prosthesis can provide functional gain, is light-weight and simple to control, the high level congenital limb deficient individual can achieve long-term success with appropriately designed prostheses (figure 4).

Figure 4: MK demonstrating active grasp.

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FUTURE LOOK OF UPPER LIMB PROSTHETICS

Lars Helmrich, Birgit Bischoff

Ottobock, Austria

INTRODUCTION

Since their development the cosmetic quality and appearance of upper-limb hand and arm prostheses has been limited. Simple hand reproductions (MyoBock Hands, Motion Control Hand, etc.) with skin-colored gloves or hook like terminal devices (Ottobock Greifer, Motion Control ETD, Hosmer, etc.) were all that was available. Also the construction of the interface was typically simple in design and shape, restricting the availability of customized designs and shapes.

In the last decade the appearance of mulitarticulating hand designs has created a new trend in technology and cosmesis with launch of the i-limb (Touch Bionics) in 2008 and the Michelangelo Hand (Ottobock) in 2010. Besides improved functionality the desired prosthetic look has changed from a natural and physiologic appearance to more futuristic presentation. Combined with various possibilities in glove and socket design, hand and arm prosthetics are presently offering individual fittings. Nevertheless, the factors for choosing a futuristic/robotic look in contrast to natural appearance still remains largely unknown.

STUDY OBJECTIVES

In the last decade the appearance of mulitarticulating hand designs has created a new trend in technology and cosmesis with launch of the i-limb (Touch Bionics) in 2008 and the Michelangelo Hand (Ottobock) in 2010. Besides improved functionality the desired prosthetic look has changed from a natural and physiologic appearance to more futuristic presentation. Combined with various possibilities in glove and socket design, hand and arm prosthetics are presently offering individual fittings. Nevertheless, the factors for choosing a futuristic/robotic look in contrast to natural appearance still remains largely unknown.

METHODS

This design analysis represents a quantitative market research of Orthotics and Prosthetics (O&P) professionals from North America, Germany, Austria, Australia, South Africa, Spain, Italy, Finland, Russia, Turkey, Poland, China, Japan, South Korea and Sweden. The questionnaire was provided to support data collection and to retrieve feedback from O&P professionals regarding their expectations of user´s preferred future upper limb prosthetic appearance**.** Possible contributing factors identified were current timeline, the users' gender, age, activity level, cultural background and the professional's knowledge of current prosthetic technology.

The questionnaire was divided into two main sets of questions. Part one targeted the O&P opinion about the prosthetic wearers´ past experience and future expectations, plus various factors that can influence the wearers' decision to use a natural or futuristic looking prosthesis. Part two gave the O&P professional the possibility to freely write about his/her expectations of the users' choice. The questionnaire was offered in paper form or as online survey. Descriptive analysis was performed on an anonymous data set via Excel.

RESULTS

The survey has been posted since November 7th, 2016 in 16 different countries. The planned availability of the survey will be until July 2017. Current results are based on 49 responses of O&P professionals. The results, given in figure 1, show the expected trends towards natural and futuristic looking prosthesis regarding: timeline, gender, age, activity level, background and the O&P knowledge of the state of the art prosthetic technology and Up-to-Date level of O&P professional (UtD). In the past, only 10% of users preferred futuristic UL prosthetic appearance, while in future this number might increase up to 51%. Correspondingly, 73% preferred a natural look in the past and 16% might prefer this appearance in the future. Females tend to prefer natural prostheses (85%), whereas men might have a tendency for futuristic prostheses (51%). 45% of younger prosthetic users (0-12) would prefer the futuristic/robotic like look, while this form of prosthesis would be chosen by 8% of people with UL deficiency older than 65 years of age. Active people have a tendency for a more futuristic look (73%) than less active people, who would rather choose a natural one (82%). Preference for a natural prosthetic appearance seems to be favored by prosthetic users with traditional background (79%). The opinion and knowledge of the O&P expert might influence the amputee´s choice to use a futuristic looking prosthesis.

DISCUSSIONS & CONCLUSIONS

According to the overall feedback from O&P professionals regarding expectations of user´s preferred future upper limb prosthetic appearance, the trend seems to indicated a more futuristic/robotic appearance, though age and gender play an important role for the individual fitting. Older users preferred a natural look, while males tend to favor futuristic prosthetic appearance.

These results reflect international O&P professionals' opinions. The feedback form actual prosthetic wearers should be acquired in the future to verify the validity of the collected data. Nevertheless, manufacturers should consider exploring this segment further.

DISCLOSURE

Authors are Ottobock employees.

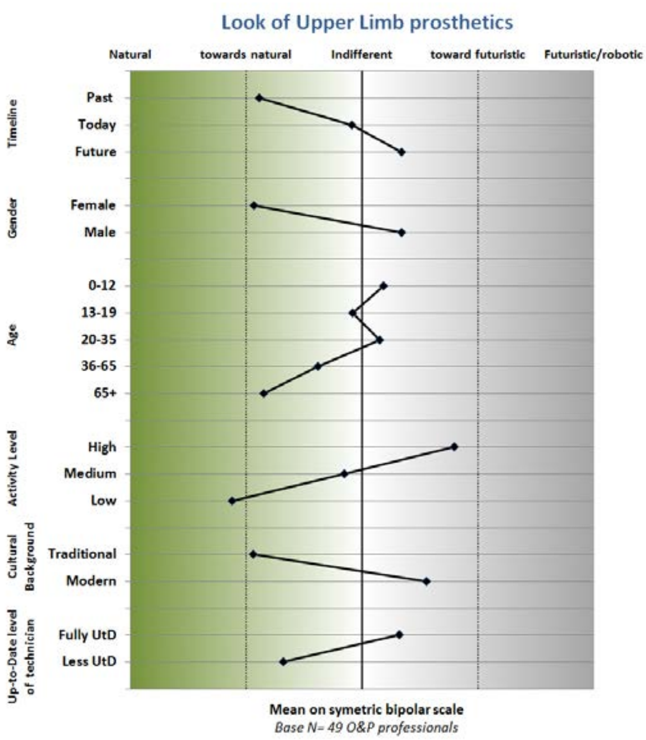

Figure 1: The future look of the upper limb prosthesis

HANDSMART GROUP
COMMUNICATION AND GLOBAL PEER SUPPORT FOR THOSE ENGAGED IN REHABILITATION OF INDIVIDUALS WITH UPPER LIMB ABSENCE

Authors for the Handsmart Group: Debra Latour[1,4], Diane Atkins[2,4], Birgit Bischof [3,4]

[1]Single-Handed Solutions, USA, [2]Department of Physical Medicine and Rehabilitation, Baylor College of Medicine, Houston, USA, [3]Otto Bock HealthCare GmbH, Austria, [4]Handsmart Group

BACKGROUND & AIM

Clinical teams in upper limb prosthetics are challenged on a daily basis due to constantly changing environment and poor education. With regard to this, the Handsmart Group was formed in February 2016 to support comprehensive clinical practice, and empower peers by creating an open access and international network.

METHOD

Sixteen international clinicians are members of the Handsmart core group and work on a voluntary basis. The larger group divided into four working groups based on their aims: enhance public awareness for the upper limb prosthetics community, establish a network among peers worldwide, support clinical practice with evidence based rehabilitation resources and acquire sources of consistent funding. Each work group discusses and votes on relevant issues to achieve their goals. The larger group will meet in person once a year to evaluate and discuss the results, methods and organization. Peers are invited to join the Handsmart network to support the group vision: Provide the most holistic rehabilitation approach for every person with upper limb loss or upper limb difference, now and in the future.

RESULTS

Based on the vision, mission and the shared core values of the group, fundamental strategic keys (Figure 1) were identified. The group developed fundamental work plans for the first year. A website (http://handsmartgroup.org) was launched in autumn 2016. This online platform supports the group to create and enlarge an international peer network, to share information, access resources and to support clinical practice for international clinicians in upper limb prosthetics.

To better accomplish the Handsmart vision and support clinical practice with evidence-based rehabilitation resources, the group will seek for evidence to support the suggested rehabilitation.

Figure 1. Handsmart Group strategy pyramid

DISCUSSION & CONCLUSION

The Handsmart core members invite external parties involved in upper limb loss/difference rehabilitation to collaborate and support the group. All initiatives will enable successful work in the promotion of its mission and will aim to improve the daily lives of clinical teams in upper limb prosthetics.

The international consortium of expert clinicians would like to thank the companies Ottobock and ProsthetiKa for their financial support and Ottobock for initiating this project. The Handsmart Group is independent and follows the international needs and interests of all people. There is no financial interest in this group. The handsmart group is currently in the process of incorporating as a 501c3, non-profit, in the United States.

DISCLOSURE

This is a work of all Handsmart Group members: Diane Atkins, Birgit Bischof, Liselotte Hermansson, Wendy Hill, Julie Klarich, Debra Latour, Ayala Nota, Sandra Ramdial, Eitan Raveh, Agnes Sturma, Shawn Swanson Johnson, Kristi Turner, Claudia Winkler, Paula Wijdenes and Daniela Wüstefeld.

IMPACT OF UPPER LIMB PROSTHESIS SIMULATORS IN PROSTHETIC REHABILITATION

Debra Latour, M.Ed., OTR/L

TRS, Boulder, CO;

ABSTRACT

Simulation technology is used in a number of health care arenas and has been utilized in the past with non-amputees as well as on the intact side of individuals with unilateral upper limb loss. This paper explores the impact of a body-powered simulator with interface to accommodate voluntary-opening and voluntary-closing terminal devices. It was used in education of occupational therapy students; in client consultation and in active therapeutic intervention.

INTRODUCTION

Evidence has been published regarding the beneficial impact of prosthesis-simulators. Bittermann (1968) cites use of such simulators with the non-amputee. This concept has been utilized for decades to impart empathy and to facilitate understanding operation of the body-powered technology. Weeks et al (2003) discusses the use of a simulator with uninvolved upper limb to successfully transfer skill of prosthesis use to the involved upper limb. Teaching individuals with upper limb deficiency to become adept with the prosthesis, its use and integration of it into acquisition of skills related to activities of daily living, work, recreation and social interactions can be challenging. As any practitioner of occupational therapy services knows, it is integral for beneficial outcomes that caregivers and other family members be involved in the process. Carryover of recommendations for all aspects of wear schedule of the prosthesis, skills-drills activities and adaptive strategies and techniques is essential for the successful outcomes of functional independence and positive perceived quality of life. Family members and other caregivers may be present during the prescriptive and therapeutic phases of the prosthetic program, but often lack first-hand experience of wearing/utilizing an actual prosthesis. Simulators of limited technology, such as a voluntary-opening device may be available to provide limited experience, but not readily accessible on an ongoing basis. This technology is typically used to provide a forecast to the consumer relative to expectations. Such simulators have also been used with clinicians and peer groups to advocate empathy and respect for individuals with UL differences and to enhance understanding of what is involved to strategically utilize body-powered prosthetic technology.

METHOD

Subjects: Subjects included distinct groups of occupational therapy students; clients, family members; and case managers. Apparatus: The VC-VO prosthesis simulators were used with each subject group for education, experience, impacting realistic expectations, providing evidence and inciting empathy. Method: Subjects were given initial training to skills drills followed by the opportunity to complete functional tasks. Sessions were recorded by video and responses were organized according to themes. Informed consent was received from human subjects. In a particular situation, a 45 year-old male who presents with acquired loss of both feet and both hands due to illness at the age of 11 months was offered the opportunity to use the simulators as pre-prosthetic preparation. His prior prosthetic experience is limited to 3 months as a child. His adaptive strategy of using both residual limbs at midline to accomplish tasks requires more time; overt posturing has appeared to cause mid and low back pain. The case study details use of the VC-VO prosthesis simulators at the time of evaluation and then weekly in pre-prosthetic training for a period of six weeks. Since delivery of the definitive prostheses the subject has engaged in prosthetic rehabilitation for additional functional skills during B-ADLs and I-ADLs. The subject was assessed using outcomes measures including the Quick DASH, Box and Blocks Test and the UNB Test of Prosthetic Function.

RESULTS

Simulators were used in the education of occupational therapy students who will likely be providers of prosthetic rehabilitation; with consumers, their caregivers/families and case managers. 100% of the individuals offered this experience stated that they better understood the demands/requirements of the technology which related to more realistic expectations of the devices. These opportunities with the simulator appear to enhance carryover of strategies to facilitate skill acquisition and appropriation of prosthetic satisfaction. Case presentations of these groups will be described during this presentation.

During pre-prosthetic phase, the client met all of his preliminary goals that included skills drills and beginning functional tasks using bilateral prostheses. At the time of delivery of his definitive prostheses, the client was able to complete many self-care tasks independently using his technology. He has since engaged in prosthetic training to refine skills toward instrumental activities of daily living including care of his young children, management of his home and property and eventual return to work. Preliminary data reflects overall satisfaction and functional ability using the definitive prostheses, and greater initial ability upon delivery due to the simulator experiences. Final outcomes will be reported at this event as the subject continues to participate in prosthetic rehabilitation.

DISCUSSION

It appears that the concept of utilizing simulators is underutilized. The body-powered prosthesis simulator described accesses both voluntary-opening and voluntary-closing terminal devices. As described in this presentation, the prosthesis simulator can be used in multiple stages of prosthetic training. At initial evaluation, it can be used to compare function and access of the technologies for successful prescription and actual client trial. This evidence can be video-taped and photographed to provide compelling evidence justifying medical necessity to the funding stakeholder(s). The caregiver can experience the diverse technologies in order to better understand the requirements of use and application to functional/bimanual manipulative tasks and case manager experience can speak to the acquisition of technology and to access to skilled therapy. During the preparatory phase, the user can adjust to the demands of suspension and practice pre-prosthetic skills drills and activities. Upon delivery of the definitive prosthesis, the simulator can be utilized to educate the family members and caregivers to various strategies in order to complete bimanual tasks. These opportunities with the simulator appear to enhance carry-over of strategies to facilitate skill acquisition and appropriation of prosthetic satisfaction.

The VC/VO prosthesis simulator was used during the pre-prosthetic delivery phase of intervention to address skills drills of grasp and release in diverse planes, functional splinter skills and bi-manual functional tasks; accompanied by work with the mirror box to occlude vision and address position in space, surface/ object feature identification and object identification. It is thought that such emphasis may help to improve functional outcomes and consumer satisfaction with the definitive prosthesis, impact user acceptance and minimize rejection of the prosthesis. This case study of the client with bilateral UL limb loss details the interventions used, reports functional outcomes, perception of ability/disability and client satisfaction of the prosthetic technology provided to him.

ACKNOWLEDGEMENTS

The Author is an occupational therapist with private practice; and has business relationships with several companies within the prosthetic industry, including the manufacturer of this technology.

Informed consent was obtained from all participants.

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PROSTHETIC USER -SATISFACTION AND CLIENT- CENTERED FEEDBACK FORM

Debra Latour, M.Ed., OTR/L
Handspring, LLC

ABSTRACT

Individuals who present with upper limb loss or congenital difference experience challenges that impact physical and psychosocial functions. Many of these individuals utilize prosthetic technology to provide or restore some of the upper limb function. As no single technology is currently available to replicate the diverse functions of the human hand, upper limb prostheses come in many forms to serve many purposes from passive assistance to complex manipulative capabilities for bimanual tasks. Today's innovative prosthetic technologies can help to restore the consumer's independent function at home, at work and in the community and improve their perceived quality of life. Existing client satisfaction tools often appear inadequate; and the information is typically requested late in the process, hampering functional outcomes and hindering the opportunity to rectify dissatisfaction. In addition to the limitations in physical function, are the impacts of prosthetic wear on self-esteem and how one performs social roles and conducts social functions. All of this can result in rejection and/or abandonment of the prosthesis. The problem is multi-dimensional and ultimately impacts all who use or might potentially use prosthetic technology.

INTRODUCTION

As healthcare professionals and providers, it is incumbent upon us to provide client-centered care. The consumer demands it, the healthcare industry requires it and our professional ethics mandate it. Scaffa, Reitz and Pizzi (2010) call us to "understand the determinants of health, such as lifestyles and living conditions, so that these can be maintained or improved". The authors cite the meaning of health as defined by the World Health Organization (WHO) to be the "complete state of physical, mental and social wellbeing, and not just the absence of disease or infirmity". Patient satisfaction has long been a buzzword and in the prosthetic industry; it includes satisfaction with service delivery as well as with technology.

Hill et al. (2009) conducted a systematic review of assessment tools relevant to this population and prosthesis use. Their findings included barriers to communication culturally, linguistically and with lack of common terminology across professions. They cited the need to implement a unified collaborative approach to improve communication between all stakeholders including the clients, clinicians and researchers (Hill et al., 2009). This viewpoint also coincides with the strategic directions of the National Prevention Strategy (2011) that include empowering people and eliminating health disparities.

The International Classification of Functioning, Disability and Health (ICF) was designed to serve several purposes such as to provide such common language and reach across the multiple health care disciplines and to provide a structure for advocacy for individuals with disabilities (WHO, 2001). The ICF model provides a framework of inter-relatedness of the health condition, environmental and personal factors to the components of body functions and structures, activities and participation. Hill et al. (2009) note that according to the ICF definitions, prostheses are perceived as assistive devices and designated as environmental factors. The disparity for individuals who utilize prosthetic technology is that for many, the prosthesis serves as an extension of the user's body. While it may serve as a tool to access bimanual functional tasks, it also becomes a replacement for the absent body structure/body function. According to Hill et al. (2009), this unfortunate classification stifles the voice of this population and ignores the experience of the prosthesis user.

Within the prosthetic industry there has been much queried about the use of the technology, how individuals perceive the technology, why they use it and how it is incorporated into the schema of the person. Few researchers have tackled and reported on the evidence as cogently as Craig Murray in his studies of 2005 and 2009. In the earlier study, he explored the factors toward adjustment and social meanings surrounding the use of prostheses and particularly sought the perceptions by limb users themselves. Several themes emerged including actual prosthesis use and social rituals, the perceptions of social isolation and the reactions of others, whether to conceal or disclose the limb difference and the social implications of each, and feelings/experiences relative to social and intimate relationships. Factors that influence adjustment and successful rehabilitation included early prosthetic fitting, prosthetic satisfaction and the need for individual expression (Murray, 2005). Satisfaction with the prosthesis is associated with increased self-esteem, increased social integration and absence of emotional challenges. The need for individual self-expression includes social expression, ‘person-first’ societal acceptance and personalizing the appearance of the prosthesis to what is perceived as aesthetically pleasing to the wearer. It is this work that served as the impetus for the development of a platform to raise the voices of the consumers and to heighten the hearing of the practitioners.

Clinicians at Handspring (based in NY with additional clinics in FL, CO, and UT) use a client-centered collaborative approach with occupational therapy and prosthetic services. They recognize the need to obtain client feedback in a systematic way that would empower the individuals and allow provision of technology that wearers of upper limb prostheses would like and would use. They collaborated with clients to create a document that uses common language and offers a feedback loop during all phases of the prescriptive prosthetic process, initiating use of the information during the pre-prosthetic phase and extending it through follow-up after delivery of the definitive technology. The form addresses specific elements of prosthesis use cited as important by the clients such as comfort of the socket, aesthetics, ease to don/doff, tolerance to weight, length, socket and harness as appropriate; control systems, reliability, pain and functionality of the technologies. The user grades each item using a 3-point color-coded system that is easy to use by children and adults. Any item that the client rates in the red column is immediately addressed during that visit; items in the yellow column are addressed subsequently. By enacting emergent practitioner response to remediate the identified problem(s), the client experiences that his/her voice has been ‘heard’, that their perceptions are important and that they as individuals are important. What first began as a client-centered feedback form to improve prosthetic satisfaction, acceptance and use has additionally and more importantly become a tool to empower the population of individuals who have experienced upper limb loss to speak and to be heard.

METHOD

Individuals with upper limb acquired loss or congenital deficiency who present for prosthetic technology are given the McGann Client Feedback Form at different stages of development of prosthetic technology and training. Additional assessment using outcomes measures including the QuickDASH, Box and Blocks Test and tests of prosthetic function as appropriate to technology developed. Data: Scores are derived from the diverse tools, correlated by subject as they relate to prosthetic satisfaction, function and self-perception of disability or quality of life. Final outcomes will be reported at this event as the subject continues to participate in prosthetic rehabilitation. This tool has been expanded to include feedback forms relative to occupational therapy and prosthetic rehabilitation, orthotic satisfaction and lower limb prosthetic satisfaction.

DISCUSSION

This presentation specifically describes the feedback form and its implementation during the prosthetic fabrication and rehabilitation process. Case studies offer insight to its correlation to scores derived from measures such as the Quick DASH and the SF-36, changes to the prostheses and impact on functional performance of the client as measured by tools such as the UNB, SHAP and ACMC.

Murray (2005 and 2009) cites the importance of consumer perceptions, input and self-advocacy to the design of prosthetic technologies. He speaks of the social meanings of prosthesis use and the value of this as it relates to user satisfaction and integration to the community. By actively engaging the client and extracting personal feedback, as well as input from the family and/or case manager, the prosthesis user is able to influence his/her care. Relationships between prosthetic satisfaction, self-perception of ability and function emerge as important facets of the rehabilitation process. As clients use their voices to note the problems they experience and the functions they enjoy, they appear to develop self-advocacy skills and to be more confident in their observations and reporting. Functional abilities appear to improve and the personal perception of ‘disability’ appears to diminish. This speaks to population health relating to occupational justice as the clients appear to be more “ability-aware”. The opportunity to provide meaningful feedback that is heard and is acted upon acts as a change agent to impact the individual consumer, the collaborative team and ultimately the care. It proves Reilly’s statement that “man through the use of his hands as they are energized by mind and will, can influence the state of his own health” (Scaffa et al., 2010).

ACKNOWLEDGEMENTS

The Author is an occupational therapist with private practice; and has business relationships with several companies within the prosthetic industry, including the prosthetic provider where this tool was developed. Participants agreed to informed consent.

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TOWARDS MORE VERSATILE GRASP: A NEW BODY-POWERED VO/VC TERMINAL DEVICE

Bradley Veatch

ToughWare Prosthetics

ABSTRACT

Terminal devices (TDs) for body-powered (BP) upper-limb prostheses typically operate using either voluntary opening (VO) or voluntary closing (VC) grasping modes. Individuals familiar with TD design recognize that under certain conditions one mode performs better than the other, and that ideally, users would be free to select—with minimal thought and disruption—the one they felt most appropriate for a given task. ToughWare Prosthetics' patented new VO/VC TD gives users this choice; the core grasping technology is purposefully robust and mechanically simple, comprising an elastic bungee-type cord, spatial lever, and contoured grasping elements. With the lever in one position, the elastic cord doubles on itself, producing a strong (additive) force used to achieve VO grasp. A second lever position causes the band to operate differentially, providing a reduced force for biasing the unit open during VC grasp. The mechanism exploits geometric symmetry between these two lever positions and the forearm cable attachment point to ensure the user's harness and cable remain correctly adjusted for optimal operation between VO and VC. Switching is accomplished by moving the spatial lever between positions; cable excursion is identical for both modes at 2-1/4 inches. Early field testing revealed that grasping contours optimized for VO operation were comparatively poor for VC, and vice-versa. In response, new contours were developed maximizing hook utility and grasping zone visual acuity for VO, and implementing a novel tilted-axis concept that optimizes grasp stability under high loads for VC that simultaneously minimizes hook interference. Replaceable compliant friction pads located on the hook faces and medial palm further enhance overall grasp quality. Designed for manufacturability, the new TD embraces state-of-the-art additive manufacturing processes in both plastic and metal to reduce cost and weight (9 ounces) while achieving an elegant, aesthetically pleasing design that is just 4-5/8 inches long. This versatile grasping technology is part of an ongoing pilot program exploring how new amputees equipped with on-demand VO and VC grasp capability employ those modes to become proficient in their use with the objective of deriving maximum benefit and enjoyment of their BP prosthetic appliances.

CASE STUDY. FITTING A UNIQUE PEDIATRIC CONGENITAL BILATERAL ELBOW DISARTICULATION

Alistair Gibson

Hanger Clinic

ABSTRACT

As clinicians we are presented with challenging and unusual cases on a regular basis.

Sometimes we draw from our experience in fitting similar cases, sometimes the case is so unique and individual that we have never experienced anything like it before, and our support network of experienced specialists have never seen before either.

I was introduced to a 5 year old overseas patient presenting with congenital bilateral elbow disarticulation, with a single long digit growth at around the level of 70% humeral length.

Surgery to provide ROM of this digit had been performed allowing a few degrees on the right side, and around 30 degrees on the left. He had been receiving excellent Occupational Therapy to increase ROM and function. I worked with his therapist to formulate a working prescription.

This presentation will outline the thought process, methodology, casting, fitting, manufacture, delivery, and follow up post delivery.

Success with prosthetic treatment at a young age is dependent on many factors, however sometimes options are limited, and will address the options of follow up care as he returns to his country.

ADVANCEMENTS IN CLINICAL APPLICATION OF CUSTOM SILICONE INTERFACE FOR PEDIATRIC PROSTHETICS

William Yule, Bill Limehouse, Branden Petersen and Patrick McGahey

Hanger Clinic

ABSTRACT

The current technology of custom silicone sockets for pediatric upper limb prosthetics has been advantageous to the pediatric patient population. Historically, pediatric prosthetic systems have been designed to meet developmental needs as a child progresses and ages with onion skin design and other flexible socket interfaces. With the advent of custom silicone socket interfaces the pediatric upper limb patient population has benefitted in more comfortable, flexible and durable myoelectric and conventional upper limb prosthetic systems.

Prosthetic practitioners have utilized various commonly accepted practices when fitting the pediatric patient. Custom silicone socket systems are now more readily available than in previous years and have presented more advantages to successful fittings than previous designs. The ultimate goal of this technology is to improve the clinical outcomes for the pediatric patient population through a better socket interface which adds comfort, flexibility and better acceptance of the prosthesis. Case studies and application will be presented to show the benefits and results applicable to this technology.

This presentation will familiarize the healthcare professional of silicone technology and its application advantages in the clinical setting as it relates to fitting this patient population. Options and methodology will be presented to educate health care practitioners as it applies to the fitting and functional applications in the clinical practice for the pediatric upper limb loss patient.

OUTCOME MEASURES IMPROVE FOLLOWING HOME USE WITH PATTERN RECOGNITION CONTROL

Levi Hargrove[1,2] , Laura Miller[1,2], Kristi Turner[1], Todd Kuiken[1,2]

[1] *Shirley Ryan AbilityLab*

[2] *Northwestern University*

ABSTRACT

Nine people with transhumeral amputations and targeted muscle reinnervation participated in a study to determine how outcome measures change pre and post a minimum 6 week home trial. Each subject controlled a prosthetic arm system comprised of commercially available components controlled using a pattern recognition control system. Subjects showed statistically significant improvements ($p<0.05$) in offline classification error, Target Achievement Control test results, the Southampton Hand Assessment Procedure (SHAP) and the Box and Blocks Test. Their performance also showed a trend toward improvement in the Clothespin relocation task and the Jebson-Taylor test; however these changes were not statistically significant.

INTRODUCTION

Pattern recognition control has been investigated for decades as an alternative to conventional amplitude control for upper-limb multifunction prostheses [1-4]. In pattern recognition control, machine learning techniques are used to decode information from residual limb muscles of the forearm[5, 6], natively innervated biceps and triceps muscles[2], or reinnervated muscle using targeted muscle reinnervation [3]. Several pattern recognition algorithms have been evaluated and many have been shown to accurately classify several movements with greater than 90% classification accuracy [7].

Most pattern recognition studies have been performed over short durations in controlled laboratory settings, often using intact limb control subjects or within virtual environments. Studies completed over multiple days show that subjects form more consistent and accurate patterns with [8, 9] or without [10] real-time control feedback. This would presumably lead to improvements controlling a physical prosthesis. We recently showed that outcome measures taken with a physical prosthesis tended to improve after using a pattern recognition control system during a 6 week home trial for transradial amputees [11].

The objective of this study was to compare a suite of outcome measures pre and post a minimum 6 week home trial. Based on the previously cited studies, we hypothesized that there would be an improvement in outcome measures as patients learned to form more consistent contractions and learned to use their physical prosthesis within their home environment.

METHODS

Nine individuals with transhumeral level amputations who had previously undergone TMR and provided informed consent were recruited for the study which approved by Northwestern University’s Institutional Review Board. Seven different surgeons performed the surgeries. All subjects were previous myoelectric prosthesis users prior to enrolling into the study, but at the time of enrollment, not all subjects were routinely using their prostheses.

The surgical method has previously been described in detail[12]. Briefly, the patients had general anesthesia with no paralytic agents so that nerves could be identified easily with stimulation. An incision was made between the two heads of the biceps. The plane between the long and short head of the biceps was identified, widened and explored to find the musculocutaneous and median nerves. The musculocutaneous nerve to the short head of the biceps was cut as it entered the muscle and the distal segment was buried in the long head so that it did not reinnervate the short head. Next the median nerve was identified and freed distally. It was then cut so that the proximal segment could be transferred to the short head motor point and the median nerve was simply sewn over the small motor point on to the muscle. For some surgeries, the subcutaneous fat was dissected free from distal to proximal and saved as a fat flap. The fat flap was then laid between the short and long heads of the biceps as a physical spacer that help to separate the EMG signals once the recovery was complete and the patient was refit with a prosthesis using TMR. Essentially this same procedure was next done to the triceps so that the distal radial nerve innervating extensor muscles below the elbow was transferred to the lateral triceps and a fat flap separated the lateral triceps from the long head and medial heads of the triceps.

A custom fabricated prosthesis was created for each patient using a Boston Digital Elbow (Liberating Technologies Inc.), a Motion Control Wrist Rotator (Motion Control Inc.), and a single degree-of-freedom terminal

Table I: Patient Demographics

Patient	Age (years)	Time since amputation (years)	Time since TMR (years)	Side	Gender	Etiology	Terminal Device used
P1	35	4	3	R	M	Trauma (military)	Hook-ETD
P2	45	2	1	R	M	Trauma (train)	Hand
P3	54	6	<1	L	M	Trauma (military)	Hook-ETD
P4	58	5	1	L	M	Sarcoma	Hook-ETD
P5	25	6	6	L	M	Trauma	Hook-ETD
P6	31	8	7	L	M	Trauma (military)	Hook-Greifer
P7	27	2	1	R	M	Trauma (crushing)	Hook-Greifer
P8	31	1	1	R	M	Trauma (MVA)	Hook-ETD
P9	44	1	<1	R	F	Trauma (infection)	Hand

device of their choice (Table 1). Consequently the prosthesis is capable of performing the following powered movements: elbow flexion (EF), elbow extension (EE), wrist pronation (WP), wrist supination (WS), terminal device open (TDO), terminal device close (TDC), and no movement (NM). Many of the terminal devices also incorporated passive wrist flexion and extension. All subjects, except P9, were fit with two custom fabricated thermoplastic elastomer gel liners (Alps Inc.). Stainless steel electrodes were embedded into the wall of the liner and stretchable conductive fabric transmitted the EMG signals to the distal end of the liner. P9 was fit with a custom rolled silicone liner to minimize length. Electrode locations were not targeted over specific muscles, rather a grid of electrodes were used as described in previous work[13]. At the distal end of the liners, the signals were amplified and digitized using a Texas Instruments ADS1299 chip sampled at 1000 Hz and transmitted to an embedded controller. The decoded commands were then sent to the prosthesis to control movement and were also logged by the embedded system so that the amount of time the prosthesis was used could be measured. This pattern recognition system was developed internally at the Center for Bionic Medicine and was subsequently released commercially as the Coapt Complete Control System (Coapt, LLC). The amplifier gains were set on a subject specific basis with a typical value of 2000, and data were digitally filtered between 70-450 Hz. A recalibration switch was laminated into the outer wall of each socket so that the users could initiate a pattern recognition calibration routine whenever they desired.

Seven of the nine subjects were naïve to pattern recognition. While the prosthesis was being constructed, these subjects were taught the concept of pattern recognition and instructed to make repeatable and distinct muscle contractions by an occupational therapist [14]. During this prehome phase of training they were given visualization exercises but received no real-time control feedback. After subjects felt that could form consistent contractions, data were collected to train and test a pattern recognition system, and 3 trials of the virtual environment based Target Achievement Control (TAC) Test were completed [15]. The pattern recognition system was identical to the system previously reported [11]. Four repetitions of 3 seconds duration for each movement were used as training data. An additional 4 repetitions of 3 seconds duration for each movement were collected following the final TAC test and were used as the testing data from which the classification error metric was computed.

After being fit with the prosthesis, subjects received intensive occupational therapy and functional use training. These sessions were spread over three or four consecutive days that lasted approximately six hours per day. Individuals took the device home for a minimum of 42 days (6 weeks) of home-use. If the prosthesis needed to be returned for repair or if the user had a valid and documented reason for not wearing a myoelectric prosthesis then additional time was added to the home-trial to ensure that they had 6 weeks of usage. Examples of valid reasons to not wear the prosthesis included extreme sports competitions, taking a beach vacation, being sunburned, etc. Outcomes were measured prior to and after each home-trial. The outcome measures included: the Southampton Hand Assessment Protocol (SHAP), the Jebsen-Taylor Test of Hand Function, three repetitions of the Box and Blocks test, and three repetitions of the Clothespin Relocation task. These measures were selected, in part, based on the recommendations of the American Academy of Orthotists and Prosthetists State of the Science meeting on Upper Limb Prosthetic Outcome Measures (25). These measures were also chosen to evaluate hand, wrist, and elbow function and were activities that could be reasonably completed with a physical prosthesis.

For outcome measures where only a single pre and post test was administered a one-tailed paired-T test was used to check differences. For outcome measures where multiple trials were recorded, a repeated measures ANOVA was performed to check for significant differences.

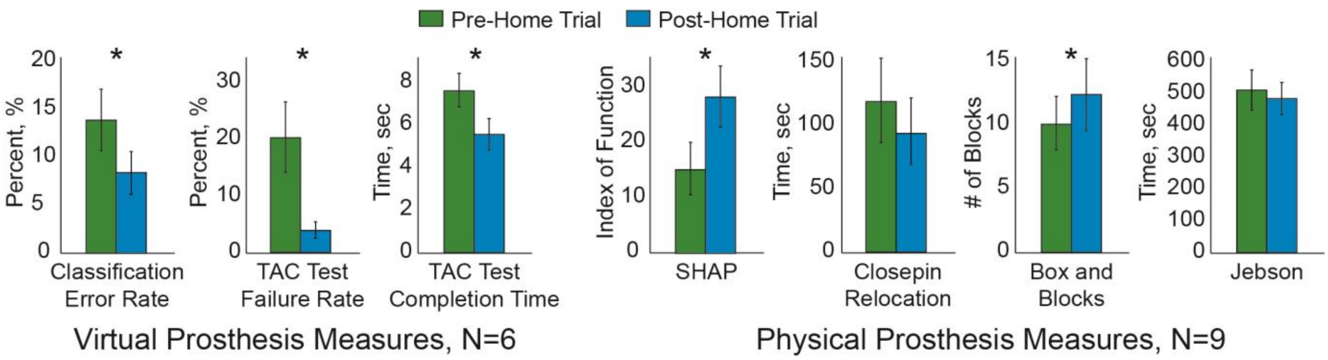

Figure 1: Outcome measures when using a virtual and physical prosthesis. The measures were taken pre and post a minimum 6 week home trial. * Denotes statistical significance between at the p = 0.05 level.

RESULTS

Classification error rate is the most frequently reported outcome measure to characterize the performance of upper-limb pattern recognition control systems. The classification error metric and TAC test outcome metric were only available from 6 of the 9 subjects. Two subjects had prior experience controlling the virtual prosthesis and data from one subject was lost due to a computer malfunction. From the remaining subjects, we found that the classification error was significantly lower (p=0.03) after the home trial. The Target Achievement Control test was completed in the virtual environment. We found that the failure rate (0<0.001), and completion times (p=0.007) were also significantly lower after the home-trial.

All subjects wore the device at home, and could successfully recalibrate the device (Table 2). Subject 2 typically removed the prosthesis while it was still powered on and it was not possible to accurately determine wear-time for this patient. Occasionally, the recalibration failed. Upon further investigation of the primary cause of these failures was a broken electrode wire.

Table II: Usage Statistics

Patient	Number of Successful/Attempted PGTs Sessions	Total Number of Days Worn	Total Wear Time in Study (hrs)
P1	7/7	9	45
P2	39/39	18	-
P3	73/77	41	181
P4	56/57	58	365
P5	10/10	36	88
P6	20/20	14	28
P7	18/18	20	127
P8	38/38	28	69
P9	60/60	32	88

All nine subjects completed the outcome measures using the physical prosthesis. All outcome measures associated with using the physical prosthesis tended to improve compared to the pre-home trial testing condition; however there were only statistically significant improvements in the SHAP (p = 0.001) and the Blocks and Box (p =0.03).

DISCUSSION

Limited previously published data has suggested that patients learn to form more consistent and distinct contractions over time, and have speculated that these would lead to improved control with a pattern recognition controlled physical prostheses. In this contribution, we have shown statistically significant improvements in the SHAP and Blocks and Box test after a minimum 6 week home trial. The results of the clothespin relocation task and the Jebson-Taylor test also showed a trend toward improvement but were not statistically significant.

Virtual environment tests are inexpensive and convenient to use. The performance metrics associated with the TAC Test showed statistically significant improvements after the home-trial. The classification error-rates achieved by the 6 subjects who completed this portion of the study were consistent classification error rates that are typically reported in the literature [3, 7, 10]. The dramatic improvement in failure rate scores and the improvement in completion time score is likely attributed to 2 factors: 1) subject had a more accurate control system that responded better to their intention, and 2) the subjects were more familiar with the TAC test itself as they had already completed the test previously. Given that the tests were spaced by at least 6 weeks, we suspect that the improvements were primarily driven by more accurate control.

The relationship between offline measures of control, such as classification error rate and real-time control performance such as those derived from outcome measures made when controlling a physical or virtual prosthesis is nebulous. Some studies report correlation [16, 17] whereas other report only a weak or no relationship [18]. Our data suggests that there is a relationship but further works need to be completed to better characterize it.

CONCLUSION

We have found that providing users with an opportunity to use a pattern recognition controller prostheses in their home-environment can result in improved outcomes. These improvements were seen in offline performance metrics, such as the classification error-rate, and real-time control outcome measures recorded when controlling a virtual or physical prosthesis. When considered with our previous work that also show improvement in outcome measures taken pre and post home trial [11], it is important to allow for adequate practice using a prosthesis prior to recommending the final control strategy.

ACKNOWLEDGEMENTS

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POWERED FLEXION WRIST WITH ELECTRIC TERMINAL DEVICE - DEVELOPMENT AND PRELIMINARY CLINICAL TRIALS

Ed Iversen, Jeff Christenson, Harold Sears, Gregory Jacobs, Scott Hosie and Tony Jacobs

Motion Control, Inc.

ABSTRACT

Recent developments in areas as diverse as TMR surgery, pattern recognition, and implantable technologies for muscle and nerve interfaces, have helped to facilitate the feasibility of practical multi-input myoelectric upper limb (UL) prostheses.

Approaching the goal of a multi-degree of freedom (DOF) prostheses, the challenge remains of dependable wrist components for wrist flexion. Components are widely used for wrist rotation – but easily utilized powered flexion is not available.

In a recent study, the kinematics of wrist rotation versus flexion was evaluated through a mathematical model (Iversen, Christenson, 2016). The kinematic analysis shows that a powered wrist flexion/extension device expands the functional workspace.

As part of a U.S. Department of Defense (CDMRP, PRORP program) effort a robust motor-driven wrist flexion component has been developed, beginning with following general targets. The summarized results are in italics:

- Compatible with myoelectric TDs – the project necessarily included new quick disconnect approaches, with the attempt to evolve a new industry standard for a rugged, high strength, and shorter q/d.

- Highly rugged –field trials show the device withstands heavy duty usage, and is water and dirt resistant.

- High torque and speed - at least 2.8 Nm torque has been attained, and may be increased.

- Light weight – a goal of 45 gm has been attained.

- High range of motion (ROM)- 80 deg. of flexion, and 45 deg. extension for both motorized and passive ROM.

- Small scale field trials – three highly active wearers (as of 2/2017) have worn the prototypes as long as four months, in daily use, helping to build the wearer data base.

The Powered Flexion Wrist developments show a positive response to the functionality of the device, specifically:

- Field trial wearers are enthusiastic about the function of the powered flexion DOF for reaching the extremes of their prosthesis ROM with the wrist and TD in a natural position, without awkward positioning of their proximal joints.

- Wearers previously using wrist rotation found that powered flexion adds greatly to function, but does not fully replace the function of wrist rotation.

- The control of multiple DOF of wrist, in a natural manner is a challenge, but existing myoelectric control may be adequate for many wearers, so that additional surgical methods will not be obligatory.

- Exchanging between more than one (or several) terminal devices also will require new hardware developments, for shorter, high strength quick disconnection.

PERFORMANCE AND SATISFACTION WITH INTUITIVE MULTIFUNCTIONAL HAND PROSTHESIS CONTROL

Sebastian Amsuess[1], Ivana Sreckovic[1], Birgit Bischof[1], Thomas Fuchsberger[2]

[1]Department of Clinical Research and Services, Otto Bock HealthCare Products GmbH, Austria

[2]Department of Hand, Plastic, Reconstructive and Burn Surgery, BG Clinic Tuebingen, Germany

BACKGROUND

Pattern recognition control (PR) functions in a different way than conventional control (CC). Instead of relying on two electrode sites to control a single degree of freedom (DoF), PR uses many electrodes and intuitive movement mapping to control several movements seamlessly.

AIM

The aim of this feasibility study is to test the performance and satisfaction of transradial amputees in prolonged home-use of PR prostheses, and to obtain feedback from certified prosthetists and trainers.

METHODS

Transradial amputees wearing prosthetic systems with CC, single opening/closing hand and active wrist rotation were enrolled in the study. Functional assessments were performed 4 times: 1) baseline with CC prosthesis-baseline, 2) 1st follow-up with the PR prostheses after fitting and training process, 3) 2nd follow-up after 1 month of PR home use, and 4) 3rd follow-up with re-fitted CC prosthesis, and consisted of performance-based (Modified Box and Blocks test (mB&B), Clothespin Relocation and Proportional Control Test) and self-reported tests (Disabilities of the Arm, Shoulder and Hand (DASH); project specific questions). The fitting and training process were rated by certified prosthetists and trainers.

RESULTS

Six patients have been enrolled in the study and fitted with the PR devices. Users were mainly male (71%), mean age 44 (± 13.4) years. Amputation etiology was trauma (100%).

All participants were satisfyingly fitted with PR prosthesis within the first visit. The fitting and training process ware rated as clear or slightly unclear with no or mild difficulty to follow the instructions.

The ability to control hand open/close and wrist rotation, measured with clothespin test, was improved with PR (transporting the clothespins from vertical to horizontal bar showed 34% improvement; from horizontal to vertical bar 18% improvement, *Figure 1*). The mB&B, was 27s (± 33.8s) prolonged at 1st follow-up and 9s (± 18.8s) at 2nd follow-up. Patients experienced mild difficulty and problems when controlling PR system. No difference was observed in DASH and the level of proportional control. 50% of participants would prefer PR over CC. Users who were already adept CC prostheses users gained disproportionally more from PR than to technically less savvy users.

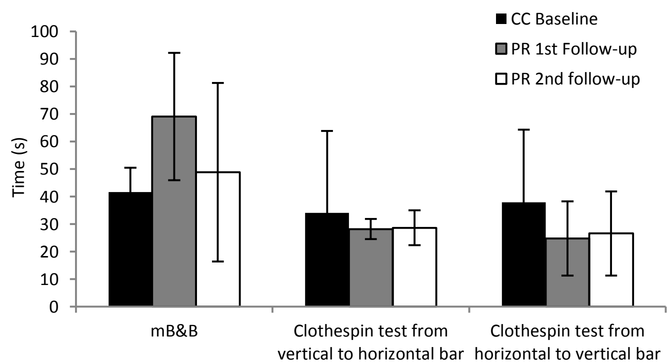

Figure1: Performance-based test conducted at baseline with CC, and at 1st and 2nd follow-up with PR.

DISCUSSION & CONCLUSION

PR improved the unilateral gross manual dexterity and ability to control two DoFs. The longer patient accommodation time and optimized product development might minimize mild problems in fine and gross motor movements observed during the first month of PR home-use.

DISCLOSURE

Sebastian Amsuess, Ivana Sreckovic and Birgit Bischof are affiliated with Otto Bock Healthcare Products.

PROSTHETIC ACCEPTANCE IN CHILDREN AND FACTORS THAT CAN INFLUENCE IT: A LITERATURE REVIEW

Ivana Sreckovic[1], Milana Mileusnic[1], Andreas Hahn[1]
[1] Department of Clinical Research and Services, Otto Bock HealthCare Products GmbH, Austria

INTRODUCTION

Children with upper limb deficiency usually do not have sense of limb-loss, and often see a prosthesis as an assistive tool, rather than a functional hand replacement[1]. Therefore a child will accept and wear a prosthesis only if it is regarded as useful[2]. The objective of this review was to evaluate the factors that might influence upper limb prosthetic acceptance in children fittings.

METHODS

Pubmed search was performed to identify the publications with upper limb prosthetic acceptance in children. The published articles additionally known to the authors were reviewed and if relevant included. Quality assessment of the studies was not conducted.

RESULTS

Eleven articles were identified as appropriate and included in the review.
41% of children were multiple prosthetic users[3]. Of those children who used only one prosthesis, 44% selected a simple passive hand as their prosthesis of choice, 41% a body-powered and 15% a myoelectric prosthesis[3]. Another study reported that 36% of children accepted a passive or body-powered prosthesis, while 38% accepted a powered hook or “pat a cake”[4]. When children transitioned to the myoelectric hands, acceptance increased to 58%[4]. The general acceptance rate of myoelectric prostheses in preschool children was 76%[5].
First fitting before 2 years of age seems to be related to higher acceptance rates[6]. 50% of children fitted at an age older than two years abandoned their prostheses compared to only 22% of children who had been fitted before the age of two years[7].
For the final type of prosthesis, children who wore an active prosthesis were more than twice as likely to wear it longer in life than children who wore a passive prosthesis[8].
Additional factors that might increase prosthetic acceptance were: prosthetic cosmetic appearance, functionality in conducting specific tasks, appropriate training and positive parental influence[9-11].
34% of tested children with trans-radial limb deficiency between the ages of 2-20 years (n=498) rejected their prosthesis[1]. The principal reasons for rejection of a prosthesis were lack of function (53% of 135 non-users), and lack of comfort (49% of non-users)[1]. Additional factors that might increase prosthetic rejection were user´s identity challenges, level of deficiency (children with higher levels of upper limb deficiency tend to wear their prosthesis longer), and negative parental influence[9,10].

DISCUSSION

The factors that drive prosthesis acceptance in children differ from those that are leading to the prosthesis rejection. Focusing on them might increase upper limb prosthetic acceptance and use later in life.

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PATIENT-SPECIFIC OPTIMUM MOTIONS: A NEED FOR MINDSHIFT IN MYOELECTRIC ONTROL OF PROSTHESES?

Ernest N. Kamavuako[1], Asim Waris[1], Irene G. Mendez[1], Kevin Englehart[2]

[1]Department of Health Science and Technology, Aalborg University, Aalborg, Denmark

[2]Institute of Biomedical Engineering, Department of Electrical and Computer Engineering, University of New Brunswick, Fredericton, NB, Canada

INTRODUCTION:

Despite the tremendous attempts in the optimization of feature sets and classifiers, the clinical usability of pattern recognition based myoelectric control has considerable room for improvement. In this study, we propose the degree of motion preference (DMP) as a step toward a patient specific optimization of motions.

METHODS

Six transradial amputees (all males, mean age 31.2 yrs.) took part in this experiment, and participated on seven consecutive days. Five to six surface bipolar electrodes were placed equidistantly about the forearm of the residual limb. Classification of the 11 motions (hand open (HO), hand close (HC), wrist flexion (WF), wrist extension (WE), pronation (PR), supination (SU), side grip (SG) fine grip (FG), agree (AG), pointer (PO); and resting state (NM)) were performed based on seven features using a linear discriminant analysis classifier. Confusion matrices for each amputee were computed. Furthermore, we investigated the best combination of six active motions plus NM per day. The optimum set was selected as the set with the highest average accuracy. Because each day may result in a different optimum set, the DMP across days was quantified as the average accuracy of each motion weighted by its occurrence frequency in the seven optimum sets.

RESULTS

Average classification error was 21.5 ± 4.3 % for all 11 motions but 25.2 ± 4.8 % for the worst combination of 6 active motions (plus rest, thus 7 motions). However, ensemble average error dropped to 5.5 ± 2.5 % using the daily optimum set of motions. Figure 1 depicts that the performance of each specific motions seems to vary across days and subjects. Results showed that DMP depends on the patient and that some motions are not preferred (Figure 2).

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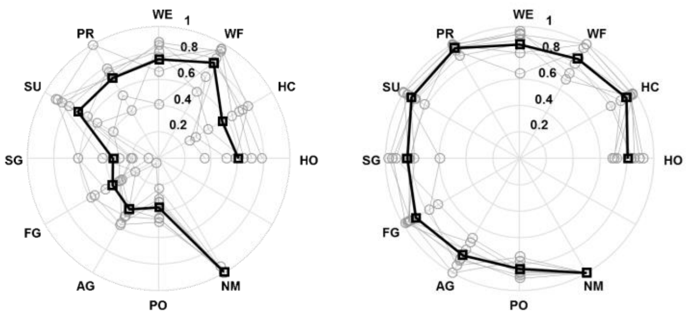

Figure 1: Diagonals of Confusion matrices of accuracies in polar form for a good (right) and a poor user (left) for each day (○) and on average (□).

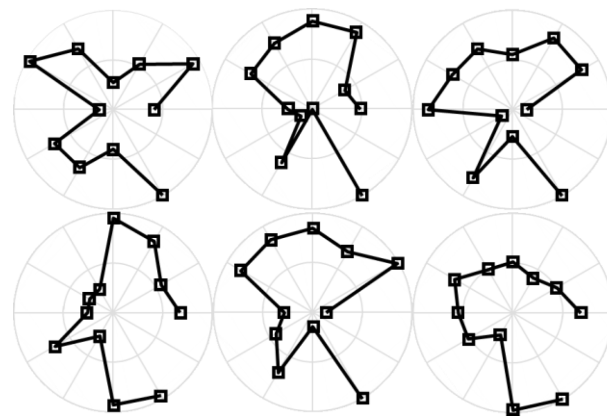

Figure 2: DMP for each amputee showing the difference in the preferred motions. Motion distribution is the same as in Figure 1, but labels are removed for visibility.

DISCUSSIONS

We have shown that selecting an optimum set of motions may improve performance; and that class performance may vary with time allowing quantification of the degree of motion preference (DMP) that is patient specific. This is clinically relevant towards patient's specific adaptive systems.

CLINICAL FITTING OF FOUR PATIENTS USING MYOELECTRIC PROSTHESES AND GEL LINERS WITH MAGNETIC COUPLING FOR TRANSMISSION OF ELECTROMYOGRAPHIC SIGNALS

Robert D. Lipschutz.[1,2]

[1]Prosthetics & Orthotics Clinical Center, Rehabilitation Institute of Chicago
345 East Superior Street, Room 1309, Chicago, IL 60611

[2]Northwestern University Prosthetics-Orthotics Center
680 North Lake Shore Drive, Suite 1100, Chicago, IL 60611

Research and development had been undertaken regarding the utilization of gel liners in conjunction with myoelectric fittings. To date; four individuals with various limb presentations and prosthetic components have been fit clinically with prostheses that incorporate gel liners and magnetic couplings that are being used for the transmission of the electromyographic (EMG) signals. The levels of amputation for these individuals include: one long transhumeral, one elbow disarticulation, and two transradial limbs. Different surgical techniques and components have been used for each of these designs based on etiology of amputation, residual limb length and patient preference. Two of the limbs, the long transhumeral and one of the transradial, have undergone targeted muscle reinnervation (TMR), while the other two have not. Three of the prostheses have: electronic wrist rotators, multi-articulating hands, and pattern recognition hardware, but not necessarily in this combination. The common factor between all of these fittings revolves around the utilization of gel liners with the magnetic coupling design.

Myoelectric fittings and sockets for upper limb prostheses have taken on many shapes and designs over the past few decades. As prosthetists attempt to create more comfortable and effective designs; the incorporation of softer interfaces has been evolving. Flexible inner sockets, such as those used in transfemoral fittings, had been proposed by Berger, et. al. in the early 1980s with the Iceland-Sweden-New York (ISNY) design. [1] Although this type of socket design, consisting of flexible thermoplastic inner socket within a fenestrated laminated outer frame, has been much of the standard in transfemoral prosthetic fittings; the popularity of this design still has yet to become a reality for upper limb prosthetic fitting. Since myoelectric (non-hybrid designs) don't require cabling for activation of segments about a joint, many prosthetists strive to provide sockets/prostheses that are "self-suspended" via the anatomy of the individual, thus eliminating the need for a harness.

Early forms of transradial, self-suspending designs include the Muenster design, with compression from anterior-to-posterior with trimlines that straddle the biceps tendon and cup over the proximal olecranon. This design is primarily used for shorter residual limbs. The Northwestern supracondylar design, with medial and lateral compression above the humeral epicondyles is used for longer residual limbs, and has a lower anterior trimline permitting greater range of motion in the sagittal plane. Another design, one created by Otto Bock, takes advantage of both anterior-posterior and medial-lateral compression and is designed for the medium length limbs. Two other socket configurations for transradial limbs, the Anatomically Contoured and Controlled Interface (ACCI) [2] and Transradial Anatomically Contoured (TRAC) [3] design had been created in the late 1990s and early 2000s. These designs incorporate many of the principles from the aforementioned three designs with added features that are claimed to benefit the wearer by providing increased stability, suspension and comfort. There tends to be debate on the overall benefits of these self-suspending designs. Tighter fitting sockets frequently make sockets more difficult to don and often decrease range of motion and comfort.

Self-suspending designs above the elbow are mainly reserved for individuals with elbow disarticulation amputations who retain their humeral epicondyles. Surgical intervention such as the Marquardt angulation osteotomy has improved suspension and rotational control for levels proximal to the elbow disarticulation. [4] Additionally, there have been more recent attempts at re-establishing humeral epicondyles by implanting hardware to replicate the shape of the amputated distal humerus. Namely, the Humerus-T-

Prosthesis has been used for both suspension and rotational control of a prosthesis with claims to have improved comfort and function for the user. [5] Various methods of osseointegration, invented by Per-Ingvar Branemark in the 1950s and improved upon in recent decades, has become a topic of great discussion as of late and has potential for revolutionizing prosthetic control for individuals with all levels of amputation, including the transhumeral level.

One of the more conventional, non-surgical means of socket alteration for the transhumeral level is based on concepts that J. Thomas Andrew, CP had provided in his review of the "Dynamic" transhumeral socket. [6] This incorporates tighter medial-lateral compression of the arm and soft tissue, with substantial anterior-posterior compression at the proximal socket, surrounding the humeral head. Although this does not provide for self-suspension, it is frequently used in conjunction with "traditional suction sockets", where the user dons the socket with the use of a donning aid, i.e. Teflon lined bag, and also provides improved control of the prosthesis over former "passive" transhumeral socket designs.

In addition to comfort within the socket, control of the myoelectric prosthesis is paramount. When the user moves through a variety of motions and tasks; traditional sockets with semi-rigid interfaces and/or packaged electrodes don't always maintain the electrode contact and control necessary to perform the intended operation. Gel liners have been used for several years in conjunction with body powered upper limb prostheses [7] and in combination with myoelectric prostheses. [8,9] Liners with "snap" or MagneSnap™ electrodes are most similar in design to those of this paper, however, it is required that the user manually attach the wires to the liner and electrode. The liner-magnet interface being described in these four prostheses is designed such that the threaded stud of the electrode dome is pierced through the gel liner from inside to out, in the location of one of the desired bi-ploar myo-sites, and is secured on the outside of the liner with a martensitic, stainless steel disk. Within the inside wall of the flexible inner socket; a countersunk, ring magnetic is recessed and mounted via a flat-head machine screw that has been inserted through the magnet and flexible inner socket with the electrode lead being secured on the outside of the socket by a washer and hex nut. (Figure 1) With this design, the user dons the liner and then the prosthesis without having to secure any of the leads to the outside of the liner or cautiously feed the liner and leads back into the prosthesis to prevent damage. As with any liner system, the orientation of the liner is critical and is made easier by the appropriate trimming of the liner, association of stitching and domes with anatomical landmarks such as scars or blemishes, and with practice.

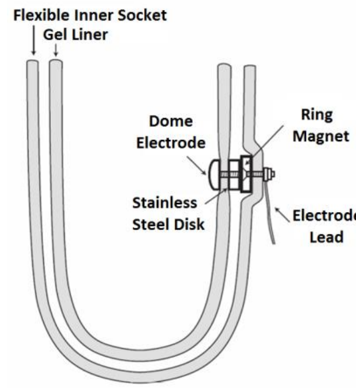

Figure 1

All of the individuals described in this manuscript have given consent to photographs and mention of their cases and prostheses. Patient 1 has a long transhumeral amputation and short transfemoral amputation secondary to trauma. He was fit with his transhumeral myoelectric prosthesis secondary to TMR procedure and involvement in research regarding the efficacy of pattern recognition control subsequent to that procedure. Although involved in research; the prosthesis was fit and provided in the clinical setting. Donning the socket using traditional suction via a donning aid proved difficult. Therefore; it was proposed to attempt fittings with the new design. Being the first individual for whom this gel liner approach was provided; several design considerations were made which have been altered as this approach has evolved. The fitting of this individual includes the customized gel liner with contacts and stainless steel disks. In addition to the contacts, a small cord had been added to the outside of the liner which aligns the liner with a recessed channel in the flexible inner socket. This feature was incorporated in order to insure appropriate alignment of the limb and liner as the socket was donned. (Figure 2) Due to the shape and length of the limb, along with all of these contacts and magnets (17 in total), this individual found it difficult to push into the socket past the magnets. A donning bag was then used as a separator between the disks and magnets to permit the limb and liner to be inserted more easily into the socket and was then removed through a traditional valve to provide both a fully seated limb and magnetic contacts between the disks and magnets. The components that were used for this


Figure 2

individual were a Dynamic Arm TMR, Electronic Wrist Rotator, Myobock Hand and Glove, and Greifer Terminal device. Pattern recognition control was used as the method for signal acquisition and processing and proved effective for all three degrees of freedom.

Patient 2 has an elbow disarticulation limb secondary to complications with Complex Regional Pain Syndrome (CRPS). His fitting took several months as he experiences significant pain in his residual limb, requiring a spine stimulator. The first approach to fitting this gentleman was using a test socket alone with harness; progressing to the addition of an endoskeletal elbow and forearm and then gradually adding weight to the device. TMR procedure was discussed, but due to the complexity of the CRPS, was not performed. The new design with gel liner was proposed and utilized due to the sensitivity of the individual's limb and difficulty in donning the socket comfortably. Donning of the liner was difficult, at first, and using the donning bag for securing the limb/liner combination into the socket was necessary early on in the fittings. As the individual became accustomed to the design and fitting, it was no longer necessary to don the device using the aid. Instead; he rolled on the liner and pushed his limb into the socket. Greater challenges for the prosthetic team were the design of the prosthesis. Because of the individual's long residual limb and the weight of the Electronic Wrist Rotator and iLimb Ultra Revolution Hand, it was decided to combine outside locking hinges with an Automatic Forearm Balance mechanism. (Figure 3) Squaring the joints, correctly contouring the proximal forearm section, placement of the pattern recognition hardware and routing of the wiring around the elbow axis proved difficult.

Figure 3

Patient 3 sustained a transradial amputation secondary to trauma and subsequently underwent TMR procedure. The time from initial fitting to completion of the device was quite lengthy due to several factors, including: reinnervation of the forearm muscles, pain in the residual limb, contralateral shoulder pain and surgery, and design of the prosthesis that was aesthetically acceptable. The latter concern was secondary to the residual limb being of mid-length and the utilization of pattern recognition hardware, battery, and *bebionic* hand. The liner system used for this gentleman has a pin locking mechanism. Although length of the overall prosthesis was of concern; the comfort of reduced trimlines and distal suspension advantages outweighed the slightly longer forearm. (Figure 4) This gentleman did not have to abduct and internally rotate his shoulder as much due to the lower trimlines and the use of a flexion wrist unit. The major concern for this user is the duration of battery life in the device. Because of this, two 2200 mAh flat, split cell batteries were combined in parallel to increase the overall capacity of the system. This has yet to prove effective in offering a full day use of the system with both Coapt pattern recognition hardware and *bebionics* hand.

Figure 4

Patient 4 had sustained quadrimembral amputations secondary to sepsis. He has bilateral transtibial, left transhumeral and right transradial amputations. Body powered prostheses had been fit to this individual and he subsequently requested a transradial myoelectric device as well. His lower limb prostheses utilize roll on liners that he dons independently. When fitting the transradial test sockets, both traditional self-suspending socket design and pin-locking liner design were attempted. This gentleman preferred the ease and comfort of donning the pin-locking liner and socket. He is able to begin to don the liner with the aid of a stationary platform or wall and using his transhumeral limb; he rolls the liner up onto his humeral section. Donning the pin locking liner with electrodes, disks and magnetic socket connection is quite easy as he has direct control (two sites) with only five disks necessary. Two pairs of disks are for obtaining the bi-polar EMG from wrist flexor and wrist extensor sites, while the fifth contact is the paired reference site. (Figure 5) The hardware being used are an electronic wrist rotator and an iLimb Quantum hand.

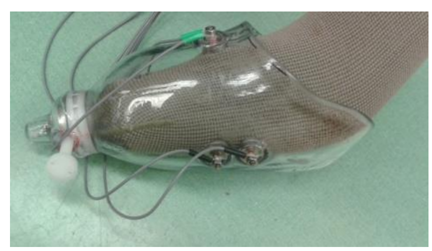

Figure 5

In four different prosthetic designs, a customized gel liner and magnetic coupling configuration have been successfully used. Whether it 5, 15, or 17 contacts, the users were able to don the devices and align the limbs such that the EMG signals were transmitted effectively through the dome electrodes, stainless steel disks, ring magnets, machine screw and wire leads down to the pre-amplifier. Utilization of liners, in this manner, improved comfort, range of motion and ease of donning as compared to traditional self-suspending designs or hardware that require fastening to the liner during donning.

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EARLY CLINICAL RESULTS OF A NEW AESTHETIC HEAVY-DUTY ELECTRIC TERMINAL DEVICE

Harold Sears, Ed Iversen, Jeff Christenson, Tony Jacobs, Scott Hosie

Motion Control, Inc., a division of Fillauer

ABSTRACT

Over the last decade, a heavy-duty Electric Terminal Device (ETD1) has been adopted widely by Upper Extremity (UE) amputees, featuring a water-resistant housing, combined with simple but functional hook fingers, motor-driven by a 2-speed transmission.

A new version was sought with goals to: 1) shorten the overall length 2) implement body-powered grip shapes to improve grip security (developed in an earlier project) [1], 3) improve aesthetics so that wearers could use a hook-style TD in a wider range of workplaces and social situations.

A new design, ETD2, using metal and plastic structure, achieves the goal of shorter length, and a smoother aesthetic, while retaining high durability, water and dirt resistance, low weight, quick response, and high pinch force, as in the legacy device. The grip surfaces are replaceable in the field, an important convenience.

The on-board electronic controller allows interchangeability with almost all other terminal devices, Bluetooth® wireless communication, and Force Limiting Auto Grasp (FLAG) [2].

The field trial subjects (n=8) were unilateral UE prosthesis wearers. Results indicate equivalent function to the ETD1 in most areas, with interesting divergence of opinion in areas. All field trial subjects signed an Informed Consent form approved by Motion Control's IRB, Ethical & Independent Review.

The usage period (from 2-18 mo.) yielded a wealth of information, guiding the design process. Summarizing the comparisons to ETD1:

- Cylindrical and flat gripping surfaces were uniformly rated superior.
- Rubber areas on lateral fingertip surfaces aided in pushing down and holding firmly, etc., for most wearers.
- Field-replaceable gripping surfaces promise to reduce the current area of highest maintenance.
- Speed and responsiveness for many was quicker than ETD1.
- Shorter overall length was valued, and produced lighter perceived weight for some.
- The aesthetics of ETD2 are appreciated, but not consistently by all. Color choices strongly favored black.
- The wider hook fingers of ETD2 meant a loss of visibility for some (but not all).

<u>Generalizations</u>

- UE prosthetic wearers as a group are enthusiastic to have more choices – as long as they do not represent a major compromise in function.
- The varieties of TD functions are different for each wearer – ensuring that opinions are very seldom consistent across all wearers.

BACKGROUND & AIMS

Over the last decade, the first generation of the heavy-duty Electric Terminal Device (ETD1) has been successfully used by thousands of UE amputees. Its success in large part may be attributed to the combination of functional hooks with a light weight motor-driven 2-speed transmission in a water-resistant housing.

Figure 1: the ETD1, using 50's era APRL hook fingers, to create a combination of simple body-powered hook shapes, with a modern motor drive in a water-resistant package.

The slender hook design provides users with the ability to reach tight places and provides high manipulation as well as visibility of objects grasped. The 2-speed transmission provides a fast closing speed and high pinch force. Water resistant housings made the ETD1 highly functional working in wet and dirty environments, e.g., the kitchen, out-of-doors occupations from auto mechanic to farming, in addition to familiar Activities of Daily Living (ADLs).

A new version was sought, with goals to 1) shorten the overall length (for equivalent length between interchangeable hand and work TDs), 2) improve grip security with wider gripping surfaces (using earlier work with body-powered TD designs), and add high-friction coatings on outside surfaces for passive functions, and, 3) allow field maintenance of the rubber gripping surfaces so highly used devices did not require frequent returns, and 4) improve aesthetics so that wearers from a broader demographic could use a heavy-duty TD in a wider range of work and social situations. All this, and importantly, retain all the functional aspects of the ETD1.

METHODS

The development process evolved a new device, the ETD2, which uses advanced integrated metal and plastic manufacturing methods contributing to an integrated aesthetic, with a strong structural core, of aluminium or optionally, steel.

The electronic features maintained from the ETD1 were:

- "Plug and play" compatibility for interchangeability with almost all other terminal devices,
- Bluetooth® wireless communication using Apple® handheld devices with an iOS operating system
- AutoCal, a built-in feature within the on-board microprocessor
- Force Limiting Auto Grasp (FLAG), an electronic method enabling the wearer to limit pinch force- which requires an internally mounted, sensitive force sensor, which at the same time is very rugged.
- A new method to allow convenient field replacement of the Gripping Pads (rubber surfaces) has been developed.
- A new splash resistant cover has been designed for the ETD2. This cover is easier to don and doff and is more aesthetic than the current system.

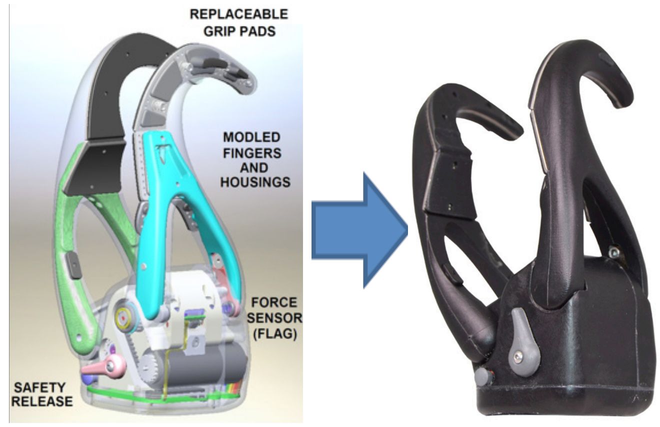


Figure 2 – The ETD2 has transitioned from the ETD1 to a shorter length, with internal structural inserts, and integrated overmolded tough exterior. Gripping Pads are replaceable in the field, for maintenance convenience.

RESULTS

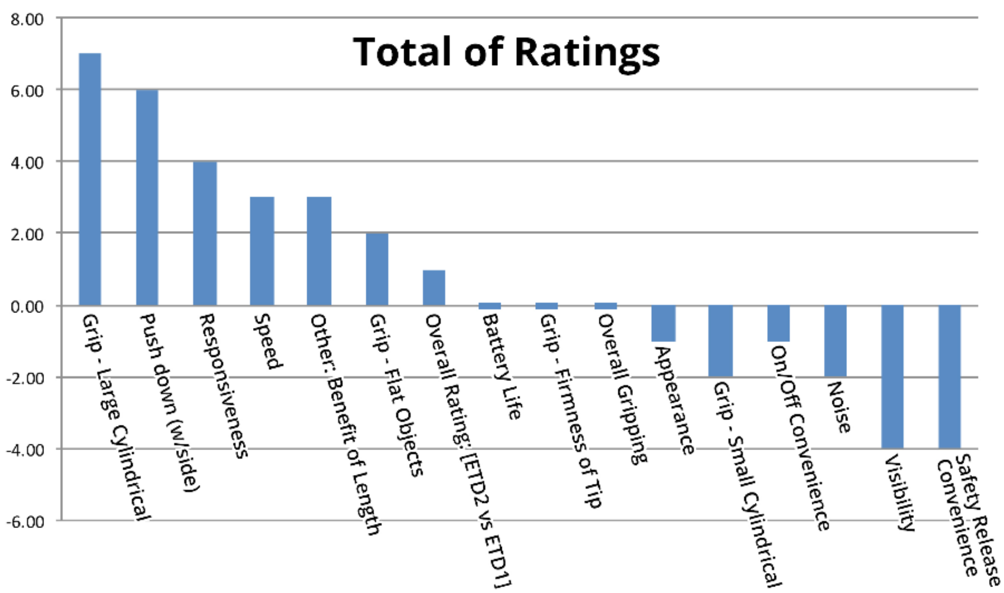
The main targets have been achieved in the ETD2 design (Figure 2). Compared with ETD1, overall length is 30 mm shorter, weight is equivalent, and strength and speed have been maintained, as has water and dirt resistance. There are two options of hook structural metal, those with lightweight aluminium inserts and those with heavy duty steel inserts.

The electronic features found in the ETD are also available in the ETD2. The FLAG feature was successfully integrated into the design.

Field Trial Results

The initial field trial (n=8) has been surveyed to obtain device feedback (Table 1).

Table 1: Summation of Ratings from field trial wearers of ETD2 (n=8). Wearers rate each feature between -2 and +2



For a quick comparison, the survey ratings are summed in Table 1, but the individual ratings were realistically evaluated for each field trial subject. For example, some subjects found the shorter length quite significant, while others did not particularly care about the length. This does not mean the shorter length was irrelevant – obviously, the importance is an individual difference. To the design team, this feature was worth the effort, especially since the shorter length was never a negative feature.

The survey results indicate improvements in some areas over the ETD1 (see Table 1). Security in gripping with the large cylindrical grip was generally highly rated. Also, greater convenience in passively pushing with outside surfaces ("Push Down w/Side"), the speed and responsiveness, flat gripping surfaces, and appreciation of shorter length all were rated positively overall. Interestingly, some subjects noted that the reduced length produces a slightly lighter perceived weight. Summation of the Overall Rating was positive for ETD2 overall (all in comparison to ETD1).

Anecdotally, from the prosthetists whose clients were in the field trials, the field-replaceable gripping surfaces generated positive feedback as well.

The survey also indicated that the visibility, noise, and small cylindrical grip ratings of the ETD2 sum slightly lower than the ETD1. Interestingly, the appearance rankings summed slightly lower than the ETD1, but again there was great individual variation, since some subjects prefer the slenderness of the ETD, despite the increased length, over the more bulky (but shorter) shape of the ETD2. Beauty, as always, is in the eye of the beholder.

Device color is another aesthetic factor, noted anecdotally. Initial field trials units were grey. However, most field trial subjects desired a different color, predominantly black. A variety of colors and/or custom coatings may be offered for the ETD2 product, when released.

CONCLUSION

ETD2 retains many of the features of the ETD1, such as rugged function, high speed and pinch force, and integration of the FLAG feature, and improves large diameter gripping, and flat grips for most wearers. In summation, ETD2 is rated slightly higher overall than the ETD1. However, hook object visibility and small diameter gripping of the ETD2 were not as functional for a few wearers.

The aesthetics of the ETD2 device are improved for some, but others find the bulky base less desirable than the slenderness of the ETD. In balance, field trials confirm the benefits of lower overall ETD2 length, and strength achieved through advanced manufacturing processes.

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POSTURAL ASYMMETRIES IN PERSONS WITH A UNILATERAL TRANSHUMERAL UPPER LIMB AMPUTATION: BIOMECHANICAL EFFECTS OF WEARING A PROSTHESIS

Takashi Nakamura and Yusuke Okita

National Rehabilitation Center for Persons with Disabilities

BACKGROUND

In persons with a unilateral upper limb amputation, postural asymmetries such as trunk rotation or scoliosis are observed. Although wearing upper limb prosthesis may be considered to mitigate the effects of limb loss on posture, few studies have been conducted regarding the influence of prosthesis during walking.

PURPOSE

The present study investigated the biomechanical influences of prostheses on the walking posture of patients with a unilateral transhumeral upper limb amputation. Kinematic analysis was used to quantify the impact of wearing different types of upper limb prostheses during ambulation, compared with not wearing an upper limb prosthesis.

METHOD

Five male patients with a unilateral transhumeral upper limb amputation (average age: 44.8 ± 16.3, average period since amputation: 2.6 ± 1.6 years) were investigated. The patients walked on a treadmill (ADAL3D-S, Medical Development) for 3 minutes (walking speed: 4.0 km/h). Patient posture was analyzed in the following five situations: 1) without a prosthesis, 2) with a socket (average weight: 200 ± 15 g), 3) with a cosmetic prosthesis (average weight: 634 ± 23 g), 4) with a body-powered prosthesis (average weight: 1220 ± 107 g), and 5) with a myoelectric prosthesis (average weight: 1600 ± 119 g). The kinematic parameters of their postures were biomechanically analyzed using 10 optoelectronic cameras (VICON, Oxford Metrix, UK).

RESULTS

The kinematics revealed that the patient's trunk rotated toward the intact side when the prosthesis was not used, and all patients swung the intact arm; the trunk slightly leaning to the intact side. When patients wore the prosthesis, the rotation and lean of the trunk decreased. The axis of the trunk rotation moved to the center of the patient's body from the intact side. The heavier the prosthesis was, the more symmetric the posture became. When wearing the prosthesis, the improved symmetry enabled increased prosthetic arm swing and decreased the trunk rotation

CONCLUSION

From the biomechanical point of view, this study showed that the patient's body posture was significantly improved when a prosthesis was used. Compensatory movements, such as abnormal swinging of the contralateral arm, were reduced. Arm swing has been suggested as a useful motion in counteracting trunk rotation in gait. The reestablishment of upper limb mass may have improved the patient's overall balance, thus, improving confidence while ambulating.

CHOOSING A MYOELECTRIC HAND AND HARDWARE THAT SUITS THE UNILATERAL AMPUTEE'S FUNCTIONAL REQUIREMENTS.

Judith Davidson

Eastern Sydney Occupational Therapy Pty. Ltd.

AIM OF THE STUDY:

Enabling the amputee to choose his own multi-functional hand has been a project for the last 4 years.

TECHNIQUES USED:

In NSW, the insurers need justification of the functional benefits of the multifunctional hand prior to its approval. It is difficult to be specific about the most appropriate hand without the use of a trial prosthesis. Since 2013 15 unilateral amputees have had trials of one or more hands prior to prescription.

The questions that are asked by all funding bodies for a prosthetic request are:

- State the participant centred goal/s that relates to this/these items of prosthesis.
- Describe why the participant needs this prosthesis. How often is this prosthesis likely to be used?
- Describe why the features/specifications of the proposed prosthesis are reasonable and necessary. Why have these components been chosen?
- Can the participant donn and doff the prosthesis independently? If not what assistance is required?
- Other information relevant to the prescription.
- What other prosthetic options / components were considered or trialed? Why are they not appropriate?

RESULTS:

Each trial costs about $5,000 if an interim socket has to be fabricated but $1,000 if they already have a suitable socket. Every insurer has approved the interim socket and trial of the hand. They can see their way to approve $5,000 without high levels of justification but the cost of $100,000 requires oversite by the NSW governing body and is much more stringent).

As a result the patients are able to determine their preference based on a variety of factors including cosmesis and function. Appropriate functional justifications dealing with specific tasks are able to be submitted for funding to easily answer the questions.

The method of prosthetic control was identified accurately, the postures that were identified, the outcome from other trials if another hand has been used. The amputee also takes responsibility for their own decisions Manufacturers know that they have to be able to loan hand to make future sales.

- MG (2015) - A partial hand amputee with intact thumb had been prescribed and fabricated a Ilimg digits prosthesis as recommended by her Solicitor. She used it well and found it useful in food preparation for 1-2 hours per day.
- JM (2014) – A farmer who suffered severe burns resulting n loss of all 5 fingers. He did not have the range of movement for any body powered prosthesis. He did not find the Limb digits strong enough for farming work and rejected them choosing to have body powered prosthesis fabricated. By that time he had developed the range of movement to find it useful.
- HL (2017) A recent referral who suffered amputation of the thumb and index of his dominant hand in late 2016. His poor English language and poor education makes his need for hand function more important. He lives alone and needs to be independent in food preparation.
- DMcS – 2011 was provided with the first Ilimb hand. He preferred the greiffer with its functionality rather than cosmesis. He was a builder by trade and worked in the Coal Mines.
- MB – 2001 who was provided with rigid grip hand. He had an intellectual disability and lived 600 km from the fitting centre. He returned in 2015 for a replacement prosthesis but chose to continue with a rigid grip hand for simplicity and ease of use.
- RMcC - 2013 He was provided with an Ilimb in 2013 which was the only multifunctional hand available. He had a body powered prosthesis which he used at work and still uses at work.
- DS (2013 and 2014) trialled the Ilimb and Michaelangelo. He did not find the 4 grips of the Ilimb easy and chose the Michelangelo for its

strength and speed. He has had a lot of difficulty with the socket and wears it occasionally.

- BW (2014)– Trialled the Ilimb and Michelangelo. However the Ilimb was a 2 week trial and the Michelangelo was only 30 minutes with a general socket. He chose the Ilimb hand but for the past 12 months it is hardly used and has spent months in repair. He states the grip is not sufficient.
- PT (2016) – He trialled both the Ilimb and Bebionic. He did not want to trial a Michelangelo. He preferred the Bebionic without a glove. He chose to utilise a roll on silicon liner which had used in cosmetic prostheses prior to this one.
- AS (2015) – He was prescribed a rigid grip hand. He wanted a trial of Bebionic and Ilimb but only had a single site control and did not have good activation of that site. We were only able to justify a rigid grip hand. He has found it difficult to make use of and progressed to a silicon liner to increase the comfort.
- RH (2016) - Over 15 years, he had used Sensor Speed Hand. He a very early Ilimb but did not like it and did not find it useful. He had a trial of Michelangelo but did not like the noise. He refused a trial of Bebionic because it did not have an electrically operated thumb for prepositioning. He chose the new Ilimb.
- RG (2013 and 2015) – He had an existing ergo elbow and internal battery. He trialled the Ilimb and while in the first week he was excited, by the end of the fortnight he stated it was too difficult to know which posture he was trying achieve. A trial of the Bebionic was successful and to date he has achieve 4 possible postures with 2 in opposition and 2 in non opposition. He wears his Bebionic hand on a daily basis.
- ML (2014) – He had undergone osseointegration before a prosthesis was trialled. He trialled both the Ilimb and Bebionic but preferred the Bebionic. The prosthetist supplied the Ilimb. He wears and uses the prosthesis intermittently. He has difficulty with skin conduction and has now undergone TMR as the first case with a surgeon in Sydney.
- TF (2016) – He has been provided with several sockets and trialled several hands. Initially he wanted the Ilimb but later on our recommendation trialled the Bebionic and eventually chose a flexion wrist. He did not want any trials but it took 2 socket designs and 3 hand trials before he had a prosthesis that was actually useful in the workshop and garden. He has an electric lock ergo elbow but uses it purely passively with his other prosthesis. A silicone liner has provided the appropriate suspension. As a short trans-humeral he would have required very proximal trim lines in his standard socket and he did not want that. .
- JC (2014) – He chose an Ilimb hand while in acute care. He wanted an electric elbow and osseointegration. Neither of these options were prescribed and he did not like the ergo elbow. He eventually chose an Ener elbow and uses it with his Ilimb hand. He has worked very hard to make it useful and wears it daily using 2-3 postures. He now wants to upgrade to an electric elbow and is consider TMR.

CONCLUSIONS

There have been 3 partial hand, 8 trans-radial and 4 transhumeral amputees.

Patients dislike harnessing and prefer silicone liners for suspension. This enable with electrode holes cut in the liners.

One partial hand did not want the Idigits due to lack of durability.

One trans-radial patient was refused multifunctional hand but approved for a rigid grip hand (due to cost)

One partial hand is awaiting approval of the trial.

12 mutlifunctional hands have been approved.

Two out of 3 transhumeral subjects have chosen to have roll on silicon liners.

. Specific tasks are identified. Not all results have been successful.

INTEGRATION OF COMFORT AND CONTROL FOR UPPER LIMB TREATMENTS

Erik Andres/ CPO, Head Department Upper Limb, Competence Center Otto Bock Healthcare GmbH, Germany

INTRODUCTION (*HEADING 1*)

According to a survey (Stark, 2013), the factors which significantly affect upper limb prosthetic acceptance, is amputation level, functional advantage, socket & harness comfort, and peer/family support.

In this presentation, the author is referring to the topics socket & harness comfort.

Since the inception of prosthetic devices usage by people with upper limb loss, the prosthetic socket design presents orthopedic technologist a challenging task. Especially because this connection- element between human body and prosthetic device must allow comfortable and secure use. For myoelectric prosthesis, an absolutely secure fit and adhesion between skin and socket is required. Every slide produces malfunctions and insecurity in the myoelectric controls. Easy donning and doffing of the device is also a very important factor for acceptance.

METHODS

People were fitted with prosthetic devices and harnesses in an individual and modern style. In these fittings, silicone use is shown with new and unique applications and approaches. Silicone is used, because of it’s well known advantages for use in medical devices.*

Successful applications presented, show use in cases from the transradial through to bilateral shoulder disarticulation level. Some examples: • Elbow-disarticulation prosthesis fitting with use of an inflatable air-bladder to increase fit through following contours • TMR with use of an integrated air- bladder to increase electrode pressure on the skin for complex control schemes • Localized use of silicone gel in a socket to provide high adhesion or to follow the contours of invaginated scars • Adaptations to individual harnesses to increase comfort and for a unique force- distribution.

RESULTS

The results presented show well accepted upper- limb sockets and harness prosthetic treatments, with outcomes showing better comfort and more long- term use of the prostheses. The applications shown will be readily usable in cases with and without myoelectric control and use available materials and techniques.

DISCLOSURE

The author is a full-time employee of Ottobock. References

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*

• Bio- compatible/ Antiallergenic

• High temperature-stability (-60 - +200°C)

• Different shore-hardness's are combinable

• Unique hygienic characteristics

• High adhesion

• Inner-layer can be coated with silicone-gel

• Durability • Highly flexible

• Positive influence on scar

The Use of Custom Silicone for a Sport-Specific Partial Hand Prosthesis: Design and 4 Month Follow-up

Kyle Sherk and Jack Uellendahl

Hanger Clinic

ABSTRACT

Prosthetic solutions for the pediatric partial hand remain custom in design and fabrication. This case study moves through the fitting and fabrication process of the prosthesis and follows the patient, AM, through 4 months of use. AM was born with a congenital limb difference of her 2nd-4th fingers of her left hand. Her right hand is normal. At age ten, she began gymnastics and came to enjoy the uneven bars. Her left hand limited her progression in the event. AM was fitted with her prosthesis in March 2016. Follow-ups continued through July 2016. This case study demonstrates that a functional, sport-specific prosthesis can be entirely of high consistency rubber (HCR) silicone.

BACKGROUND

Prosthetic solutions for the partial hand and partial finger levels of amputation continue to expand. However, specific designs for sport application have not increased for these populations and the size of many of these devices are too large for pediatric application.

PATIENT

The patient, AM, is an 11 year old girl, weighing 65# (29.5 kg) and standing 4' (1.22m) tall. She has a congenital partial aphalangia of the left hand (2nd-4th fingers). Her right hand is normal. She was beginning to work the uneven bars in gymnastics, but was limited by the grip of her left hand. This patient desired to continue advancing in her skill level. A typical partial hand solution of a wrist-based prosthesis was not an acceptable solution as it would limit her wrist motion, which can be crucial to acceleration and deceleration during the exercise.

PROSTHESIS

A custom, all silicone oppositional prosthesis was fabricated to restore the finger lengths of AM's left hand. All of the minor fingers were encapsulated within a low durometer (Shore ~25) custom silicone interface. The device restored power grip (flexed fingers 2-5) [1]. To increase the durability of the device, higher durometer HCR silicone (Shore ~80) was utilized for the finger extension. The final contour of the finger extension was directly shaped around a bar of approximate diameter corresponding to the uneven bars.

Figure 1: The low durometer interface encompassing the minor fingers.

FOLLOW UP

AM was successful in increasing her skills and speed on the un-even bars with the new prosthesis within a week. She soon began producing sufficient centripetal force around the bar, that the prosthesis would inadvertently doff and cause her to fall. A double ring closure Dacron strap was added to the prosthesis at the wrist to increase the hold of the prosthesis to AM's hand. This was initially successful but soon proved insufficient to retain the prosthesis in place as AM increased her speed around the bars. Velcro was added to the strap to increase the security of the closure in April of 2016. The last report (July 2016) from the patient's mother was of gratitude for our work on the prosthesis: AM had placed 9th on the uneven bars in a gymnastics meet against able-bodied adolescents.

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LINEAR, KURTOSIS AND BAYESIAN FILTERING OF EMG DRIVE FOR ABSTRACT MYOELECTRIC CONTROL

Matthew Dyson[1], Jessica Barnes[1] & Kianoush Nazarpour[1, 2]

[1]*School of Electrical and Electronic Engineering, Newcastle University, UK*

[2]*Institute of Neuroscience, Newcastle University, UK*

ABSTRACT

Introduction:

Signal processing of sEMG is currently the most prevalent method used to control active hand prostheses. For control purposes sEMG is typically transformed into a feature space representation prior to presentation to a controller or classifier. In this study we compare three signal processing techniques for myoelectric control based on low level EMG contractions: mean-absolute-value (MAV), a Bayesian estimate of the EMGs 'neural drive', and sequentially updated real-time Kurtosis.

Method:

Tests were performed while ten participants learned to control an abstract myoelectric-controlled interface (MCI). EMG was recorded from the abductor pollicis brevis (APB) and the abductor digiti minimi (ADM). Features were calculated over a 750 ms window and updated continuously. Participants used isometric muscle co-contraction to control the position of a 2-D cursor toward pseudo-randomly presented targets. Cursor position was determined solely by muscle activation estimates. After each trial participants received a score indicating how well they were able to hold the cursor within the target area. Participants performed 4 blocks of 72 trials. Trials were visually inspected for artifacts, all artifact free trials were used in further analysis.

Results:

The Linear filter outperformed the Kurtosis based method and produced similar percent hold rates to the Bayesian method tested, as shown in Figure 1. Rates of improvement in overall score were more apparent in the Bayesian and Kurtosis filters, while improvement in target hit rate was largely restricted to the Bayesian filter. Differences in hit rate may have been attributable to the Kurtosis and Linear methods being sufficiently similar for participants to generalise between the two. In contrast, optimal co-contraction behaviour for the Bayesian filter is likely to be different as was the rate at which the cursor moved.

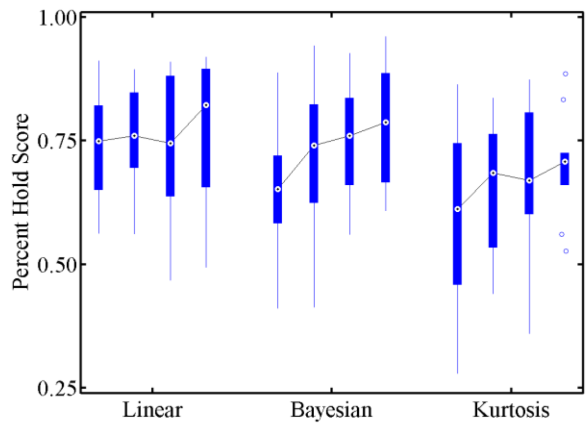

Figure 1: Percent hold score over runs for the Linear, Bayesian and Kurtosis filters.

Despite significantly less efficient trajectories, the Bayesian filter showed a reduced time required to reach individual targets. Participant performance was analysed with respect to Fitts' Law. Analysis showed decreasing accuracy relative to speed for each filter type. A significant correlation found between participant accuracy and overall cursor speed for the MAV filter, suggesting participants were able to make a speed accuracy trade-off. These results suggest that the slower pace of cursor feedback provided by the MAV filter more readily allows for adaptation in participants control strategy.

Conclusion:

Results demonstrate that linear methods can outperform more complex filtering techniques, and that real-time kurtosis may be used as an activation estimator.

PRE-CLINICAL APPLICATION OF MUSCLE SYNERGIES FOR ABSTRACT MYOELECTRIC CONTROL

Matthew Dyson[1] & Kianoush Nazarpour[1, 2]

[1]*School of Electrical and Electronic Engineering, Newcastle University, UK*

[2]*Institute of Neuroscience, Newcastle University, UK*

ABSTRACT

Introduction:

Prosthetics are often controlled by the surface electromyogram (sEMG) of muscles remaining in the limb to which the prosthesis is attached. A major challenge for next-generation prosthetics is to design a proportional control scheme that allows for control of a number of degrees of freedom. Muscle synergies are coordinative structures which act as discrete low-level units typically combined to construct a diverse range of physical movements. In this work we compare use of two channel sEMG against multiple weighted electrodes, representing muscle synergies, in control of an abstract myoelectric user interface (MCI) capable of proportional control.

Method:

sEMG was recorded from the forearm in transradial amputee participants and the upper arm in a participant with a transhumeral congenital deficiency. Participants used isometric muscle contractions to operate a MCI which displayed a cursor on the screen. Users were assessed in their ability to use the MCI under two conditions; using an optimal pair of electrodes and using a weighted combination of eight electrodes representing synergistic muscle output. Muscle synergies were derived using a data driven approach based on Principal Components Analysis (PCA). Cursor position was determined solely by sEMG muscle activation in two simultaneous channels or two simultaneous principal components. Participants began using a four target interface and progressed to eight targets where appropriate. A 1.5 (s) trial was composed of two 750 (ms) periods, the first for movement and the second for holding the cursor in the target. Percent hold score was calculated based on the duration for which the cursor was within, or in contact with, the target during the hold period.

Results:

Overall final trial scores achieved using PCA weighted electrodes were significantly higher than those based on two channel sEMG when using the four target interface. Similarly, cursor path efficiency when using the four target interface was significantly higher for PCA based control than use of sEMG. Final scores for the eight target interface were greater using PCA weighted electrodes however additional data is required before significance testing.

Conclusion:

Initial results demonstrate that amputee participants can learn to use arm and forearm muscle pairs to flexibly control the position of a myoelectric cursor in a 2-D task environment. Data-driven spatial weighting of sensors generally produced more reliable results than a single pair of electrodes. Improved scores maintaining the cursor within target sectors are mirrored in improved cursor path efficiency.

REAL-TIME EVALUATION OF DEEP LEARNING-BASED ARTIFICIAL VISION FOR CONTROL OF MYOELECTRIC HANDS

Ghazal Ghazaei[1], Patrick Degenaar[1, 2] and Kianoush Nazarpour[1, 2]

[1]*School of Electrical and Electronic Engineering, Newcastle University, UK*

[2]*Institute of Neuroscience, Newcastle University, UK*

ABSTRACT

The loss of any limb, particularly the hand, affects an individual's quality of life profoundly. An artificial arm, or prosthesis, is an example of technology that can be used to help somebody perform essential activities of daily living after a serious injury or health condition that results in the loss of their arm. Assistive technology solutions augmented with computer vision can enhance the quality of care for people with sensorimotor disorders. The goal of this work was to enable two trans-radial amputees to use a simple, yet efficient, computer vision system to grasp and move common household objects with a two-channel myoelectric prosthetic hand. We developed a deep learning-based artificial vision system to augment the grasp functionality of a commercial prosthesis. A convolutional neural network was trained with images of over 500 graspable objects. For each object, 72 images, at 5˚ intervals, were available. Objects were grouped into four grasp classes, namely: pinch, tripod, palmar wrist neutral and palmar wrist pronated. We implemented the proposed framework in studies involving two trans-radial amputee volunteers to control a commercial i-limb Ultra™ prosthetic hand and a Motion Control™ prosthetic wrist. After training, subjects successfully picked up and moved the target objects with an overall success of about 88% in various visual feedback conditions. The use of a deep-learning based computer vision system has the potential to enhance the functionality of the upper-limb prostheses in clinic.

ENVIRONMENTAL BARRIERS TO PARTICIPATION AND FACILITATORS FOR MYOELECTRIC PROSTHESIS USE- A COMPARISON WITH USERS OF OTHER ASSISTIVE TECHNOLOGY

Cathrine Widehammar[1,2], Helene Lidström[3], Liselotte Hermansson[2,4]

[1]*Department of Pediatrics, Faculty of Medicine and Health, Örebro University, Örebro, Sweden*

[2]*University Health Care Research Center, Faculty of medicine and health, Örebro University, Örebro, Sweden*

[3]*Department of Social and Welfare Studies, Faculty of Medicine, Linköping University, Linköping, Sweden*

[4]*Department of Prosthetics and Orthotics, Örebro University, Örebro, Sweden*

INTRODUCTION

Myoelectric prostheses (MEP) are used in varying degrees; the number varies between 12-80%[1]. Prosthesis use is greatly affected by the environment, and qualitative research implies that the experience from environmental influence differs depending on how much the MEP is used; daily prosthesis users experience more support and less environmental barriers [2]. To strengthen this conclusion and also to investigate if it is valid for other types of advanced assistive technology (AT), a further study based on quantitative methodology is needed.

AIM

To describe the presence of environmental barriers to participation, and facilitators for MEP use, and to compare this with users of powered mobility devices (PMD) and assistive technology for cognition (ATC).

METHOD

A cross-sectional survey was conducted with users of MEP, PMD and ATC. The inclusion criteria were: at least one year experience as AT user; age 20-90 years; and communicating in Swedish. The survey contained the Swedish version of Craig Hospital Inventory of Environmental Factors (CHIEF-S) and a study-specific questionnaire focusing on facilitators for AT use. The sample consisted of 156 participants (users of MEP n=51; PMD n=56; and, ATC n=47). The experience of using AT varied between 1-41 years, and many participants used their AT daily (MEP= 80%, PMD=64%, and, ATC=87%). Since the scores were not normally distributed, Kruskal Wallis and 2-tailed Mann-Whitney U test for differences between the groups, and Spearman's rank order correlation were used for analyses.

RESULTS

The top two items acting as barriers to participation in MEP users were *Natural environment* (temperature, terrain and climate) and *Policies government* (rules, regulations governed by law). Barriers to participation were significantly less for MEP users than for users of other AT (CHIEF-S total score, md: MEP=0.120, PMD=0.619, ATC=1.560 [$p<0.05$]). In contrast to other AT use, a significant ($p<0.05$) correlation between prosthesis use and barriers to participation was shown in MEP users, with less barriers correlating to more use. Most support came from *Related persons* and *Professionals*, and least from *Authorities* and *Rules and regulations*.

DISCUSSION & CONCLUSION

The results confirm earlier qualitative research but show a difference to users of other AT. This should be an avenue for future research. Furthermore, prosthesis usage reported in this study was higher than in most other studies. Hence, the results may not be representative for MEP users in other contexts and this need to be studied further.

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3D-GAZE AND MOVEMENT: A NOVEL METRIC OF VISUAL ATTENTION TO MEASURE UPPER LIMB PROSTHETIC FUNCTION

[1,2]Jacqueline S. Hebert, [1]Ewen Lavoie, [1] Quinn Boser, [1]Aïda M. Valevicius ,[1,2]Albert H. Vette, [1]Patrick M. Pilarski, [1]Craig S. Chapman

[1] University of Alberta, Edmonton, Alberta, Canada
[2] Glenrose Rehabilitation Hospital, Alberta Health Services, Edmonton, Alberta, Canada

ABSTRACT

A lack of sensory feedback is often highlighted as a limiting factor to the use of powered upper limb prosthetic devices. From a functional perspective, sensory feedback is hoped to reduce cognitive burden and lessen the need for visual attention to task. However, standardized methods to quantify visual attention are limited, and there is a large technological burden with certain methods. We developed a novel method of integrating eye and motion tracking during defined functional tasks, which allows identification of eye gaze behaviour in relation to object interaction and body kinematics for each segment of movement.

Twenty able-bodied participants with normal upper limb function and 10 prosthetic users participated in the study. They performed two functional upper limb tasks, a pasta box task and a cup transfer task, with synchronized upper limb motion capture and eye tracking data collection. Using the kinematic data, each movement was segmented into 4 movement phases: Reach, Grasp, Transport, and Release, and eye fixations were analyzed according to: current location being acted on by the hand (‘Current’), the future location that the hand will act on when it has completed its current action (‘Future’), and the hand itself when no other AOI is being fixated (Hand). The results of one transhumeral prosthetic user, using two different prosthetic devices compared to normative results, is presented.

The prosthetic user showed significantly slower movement times, spending a disproportionally longer time in the *grasp* phase compared to the other phases. With respect to eye behaviour, in the *reach* phase, the prosthetic user spent relatively more time fixated particularly on the myoelectric terminal device compared to the body powered condition, with both prosthetic conditions having a greater ratio of fixation time to hand than normative subjects. During *transport* of the object, the ratio of eye fixation to hand was dramatically increased in both prosthetic conditions compared to normative. This resulted in the prosthetic users spending relatively less time during transport fixating on the target to where they were moving the object (ie. reduced “look ahead” fixations).

This novel method of eye gaze behaviour analysis precisely quantified the visual attention demands of a prosthetic user during functional tasks, and created a normative data set for comparison. There were some interesting differences between myoelectric and body powered prosthetic performance that may have been due to proprioceptive feedback. Measuring changes in visual demand with new interventions may give insight into the relative importance of vision in accomplishing tasks for prosthetic users.

ACKNOWLEDGEMENTS

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IMPROVING HAND AND WRIST ACTIVITY DETECTION USING TACTILE SENSORS AND TENSOR REGRESSION METHODS ON RIEMANNIAN MANIFOLDS

Noémie Jaquier[1], Claudio Castellini[2] and Sylvain Calinon[1]

[1]*Idiap Research Institute, Martigny, Switzerland*
[2]*DLR - German Aerospace Center, Wessling, Germany*

ABSTRACT

Simultaneous and proportional control of a prosthetic hand and wrist is still a challenging issue, although giant steps have lately been made in this direction. In this paper, we study the application of a novel machine learning method to the problem, with the aim to potentially improve such control. Namely we apply different kernels for tensor Gaussian process regression to data obtained from an advanced, flexible tactile sensor applied on the skin, recording muscle bulging in the forearm. The sensor is a modular, compact bracelet comprising 320 highly sensitive elements organized as a tactile array. The usage of kernel functions with tensor arguments and kernel distances computed on Riemannian manifolds enables us to account for the underlying structure and geometry of the tactile data. Regression accuracy results obtained on data previously collected using the bracelet demonstrate the effectiveness of the approach, especially when using Euclidean distance and Kullback-Leibler divergence-based kernels.

INTRODUCTION

Despite recent advances in externally-powered prosthetics, intuitive and robust control of polyarticulated prosthetic hands and wrists by amputees remains an unsolved problem, mainly due to unadapted user interfaces and inadequate sensorization [1]. Despite remarkable advanced in this direction, e.g., [2, 3, 4], a full, clinically accepted application still has to appear. The recent results of the ARM competition of the Cybathlon, won by Robert Radocy of TRS Prosthetics wearing a body-powered prosthesis[1], stand as a powerful warning for the scientific community.

In this paper, we advance the usage of *tactile sensing* or tactile myography (TMG) to detect hand and wrist movement in a non-invasive way in order to replace or augment the traditional surface electromyography (sEMG). We bring the qualitative analysis of TMG data performed in [5] one step further, by studying regression methods allowing to account for the structure and the geometry of the muscle bulging data. This is enforced using *tensor Gaussian Process Regression*, a technique consisting of predicting a posterior Gaussian density for new inputs knowing a set of input and output data.

We first describe the proposed regression method, then we quickly review the experimental setup and data collection process, and lastly we present our experimental results. A quick discussion completes the paper.

PROPOSED APPROACH

Mathematical background

Gaussian Processes (GP) are a class of probabilistic models that defines a posterior over functions given a set of input and output data. The distribution is assumed to be Gaussian with some mean and covariance. The covariance is computed using a kernel function as a measure of similarity. The idea behind GP is that if two input points are similar according to the kernel, the output of the function at those points will also be similar [6].

Tensors are generalization of vectors and matrices to higher dimensions. They provide an efficient and natural way to represent structured multidimensional data such as videos sequences or electroencephalography (EEG) data. Recently, regression methods have been extended to tensor data, allowing an efficient exploitation of their structure, see for example [7, 8].

Riemannian manifolds are mathematical spaces that locally resemble a Euclidean space. A Riemannian manifold is a smooth manifold whose tangent space is equipped with an inner product [9]. Such model conserves the underlying geometry of the data. Examples of well known manifolds are the surface of hyperspheres (to represent orientations), the space of symmetric positive definite (SPD) matrices [10], or the space of p-dimensional subspaces in a n-dimensional Euclidean space called the Grassmann manifold [11].

Tensor Gaussian Process Regression

Given a dataset of N observations $\{(\boldsymbol{\mathcal{X}}_n, \boldsymbol{y}_n)\}_{n=1}^N$, concatenated as $\boldsymbol{\mathcal{X}} \in \mathbb{R}^{N\times I_1\times\ldots\times I_m}$ and $\boldsymbol{Y} \in \mathbb{R}^{N\times J}$, we are interested in making prediction for a new input $\boldsymbol{\mathcal{X}}_*$. By extending GP regression to tensor inputs, the predictive dis-

[1] see http://www.cybathlon.ethz.ch/de/cybathlon-news/resultate/arm-resultate.html as well as http://www.trsprosthetics.com.

tribution of $\boldsymbol{y}_*$ corresponding to $\boldsymbol{\mathcal{X}}_*$ can be inferred as

$$\boldsymbol{y}_*|\boldsymbol{\mathcal{X}}_*, \boldsymbol{\mathcal{X}}, \boldsymbol{Y}, \boldsymbol{\theta} \quad \sim \quad \mathcal{N}(\overline{\boldsymbol{y}}_*, \text{cov}(\boldsymbol{y}_*)), \tag{1}$$

where

$$\begin{aligned} \overline{\boldsymbol{y}}_* &= k(\boldsymbol{\mathcal{X}}_*, \boldsymbol{\mathcal{X}})\big(k(\boldsymbol{\mathcal{X}}, \boldsymbol{\mathcal{X}}) + \sigma^2 \boldsymbol{I}\big)^{-1} \boldsymbol{Y}, \\ \text{cov}(\boldsymbol{y}_*) &= k(\boldsymbol{\mathcal{X}}_*, \boldsymbol{\mathcal{X}}_*) \\ &\quad - k(\boldsymbol{\mathcal{X}}_*, \boldsymbol{\mathcal{X}})\big(k(\boldsymbol{\mathcal{X}}, \boldsymbol{\mathcal{X}}) + \sigma^2 \boldsymbol{I}\big)^{-1} k(\boldsymbol{\mathcal{X}}, \boldsymbol{\mathcal{X}}_*), \end{aligned} \tag{2}$$

and $(\boldsymbol{K})_{ij} = k(\boldsymbol{\mathcal{X}}_i, \boldsymbol{\mathcal{X}}_j)$ is the covariance or kernel matrix [12, 13].

Kernels with tensor inputs on manifolds: In order to exploit both the structure and the geometry of tensor inputs, the kernel is defined as a product of M positive semi-definite factor kernels

$$k(\boldsymbol{\mathcal{X}}, \boldsymbol{\mathcal{X}}') = \prod_{m=1}^{M} k(\boldsymbol{X}_{(m)}, \boldsymbol{X}'_{(m)}), \tag{3}$$

where $\boldsymbol{X}_{(m)} \in \mathbb{R}^{I_m \times I_1 I_2 \ldots I_M}$ is the mode-m matricization or unfolding of tensor $\boldsymbol{\mathcal{X}}$ [12, 13]. Each factor kernel measures the similarity between mode-m unfolding of two tensors. We consider factor kernels in the form of Radial Basis Function (RBF) kernels defined as

$$k(\boldsymbol{X}_{(m)}, \boldsymbol{X}'_{(m)}) = \exp\left(-\frac{d(\boldsymbol{X}_{(m)}, \boldsymbol{X}'_{(m)})}{2\beta_m^2}\right), \tag{4}$$

where $d(\boldsymbol{X}_{(m)}, \boldsymbol{X}'_{(m)})$ is a distance measure. Kernels based on different distances for matrices are presented below.

Kernels based on Kullback-Leibler divergence: The Kullback-Leibler divergence measures the difference between two probability distributions p and q. In our case, each $\boldsymbol{X}_{(m)}$ is treated as a Gaussian generative model with I_m variables and $I_1 I_2 \ldots I_M$ observations and parameters $\boldsymbol{\mu}_m$ and $\boldsymbol{\Sigma}_m$ [12]. The corresponding probabilistic distance measure is defined as

$$d_{\text{KL}} = \text{KL}\Big(p(\boldsymbol{X}_m|\boldsymbol{\mu}_m, \boldsymbol{\Sigma}_m) \| q(\boldsymbol{X}'_m|\boldsymbol{\mu}'_m, \boldsymbol{\Sigma}'_m)\Big), \tag{5}$$

where the Kullback-Leibler divergence between two Gaussian distributions $\mathcal{N}_0(\boldsymbol{\mu}_0, \boldsymbol{\Sigma}_0)$ and $\mathcal{N}_1(\boldsymbol{\mu}_1, \boldsymbol{\Sigma}_1))$ is $\frac{1}{2}\big(\text{tr}(\boldsymbol{\Sigma}_1^{-1}\boldsymbol{\Sigma}_0) + (\boldsymbol{\mu}_1 - \boldsymbol{\mu}_0)^\top \boldsymbol{\Sigma}_1^{-1}(\boldsymbol{\mu}_1 - \boldsymbol{\mu}_0) - k + \ln \frac{\det \boldsymbol{\Sigma}_1}{\det \boldsymbol{\Sigma}_0}\big)$ [14].

Kernels on the manifold of SPD matrices: Different metrics on the manifold of SPD matrices $\mathcal{S}_{++}^n$ may be used to define positive definite kernels [15]. The log-Euclidean metric $\| \ln \boldsymbol{\Sigma}_0 - \ln \boldsymbol{\Sigma}_1 \|_{\text{F}}$ corresponds to the geodesic distance between two SPD matrices $\boldsymbol{\Sigma}_0$ and $\boldsymbol{\Sigma}_1$, e.g. the shortest path between two elements on the manifold. It yields the corresponding distance

$$d_{\text{logSPD}} = \| \ln(\text{cov}(\boldsymbol{X}_m)) - \ln(\text{cov}(\boldsymbol{X}'_m)) \|_{\text{F}}, \tag{6}$$

where $\text{cov}(\boldsymbol{X}_m) \in \mathbb{R}^{I_m \times I_m}$ is the covariance matrix of $\boldsymbol{X}_m$. Similarly, we exploit the non-geodesic metric $\|\boldsymbol{\Sigma}_0 - \boldsymbol{\Sigma}_1\|_{\text{F}}$ yielding the distance

$$d_{\text{SPD}} = \|\text{cov}(\boldsymbol{X}_m) - \text{cov}(\boldsymbol{X}'_m)\|_{\text{F}}. \tag{7}$$

Kernel on the Grassmann manifold: The Grassmann manifold $\mathcal{G}_{n,p}$ is the space of p-dimensional subspaces in a n-dimensional Euclidean space. In this manifold, it is not possible to find a geodesic distance that yields a positive definite kernel [15]. We use the non-geodesic projection or Chordal metric $\|\boldsymbol{Y}_0\boldsymbol{Y}_0^\top - \boldsymbol{Y}_1\boldsymbol{Y}_1^\top\|_{\text{F}}$, where $\boldsymbol{Y}_0, \boldsymbol{Y}_1 \in \mathcal{G}_{n,p}$ to define the projection Gaussian kernel or Chordal distance-based kernel with

$$d_{\text{Chordal}} = \|\boldsymbol{V}_m\boldsymbol{V}_m^\top - \boldsymbol{V}'_m\boldsymbol{V}_m'^\top\|_{\text{F}}. \tag{8}$$

Here, $\boldsymbol{V}_m$ corresponds to the right orthonormal vectors of the SVD decomposition of the mode-m unfolding $\boldsymbol{X}_m$.

EXPERIMENT

In order to test the applicability and accuracy of the presented technique, we applied it to the dataset presented in [5]. We give here a very short description of the materials and methods used, see the original paper for details.

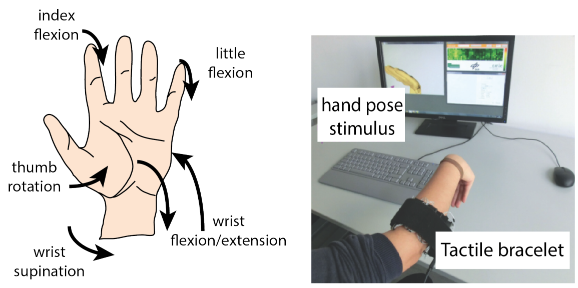
Experimental setup

The device used to capture muscles bulging around the full circumference of the arm is a shape-conformable tactile bracelet. The bracelet uses high-performance resistive elastomer-based tactile sensor technology, built upon the fact that the interface resistivity between two electrodes changes according to the applied load. The layout of the bracelet is such that a total number of 320 pressure sensors, arranged in a 8×40 torus shape around the forearm, gather a high-spatial-resolution (5mm) "pressure image" exerted by the deformation of the muscles engaged in moving the hand and wrist. The idea in itself is well-known, initially invented and studied by Craelius and others [16, 17], and its effectiveness is being studied with remarkable results, even when tested on amputated subjects [18]. TMG can be seen as a high-resolution version of force myography as laid out in the mentioned papers; Radmand et al. [19] provide a striking example of TMG applied to intact subjects, showing excellent classification accuracy. In order to provide a better form of simultaneous and proportional control, we hereby focus upon regression instead of classification.

In the experiment reported in [5], the ground truth was obtained by simply using the values of an animated visual stimulus, namely a hand model with nine degrees of freedom, with the understanding that the subjects would follow it with reasonable accuracy. This is an instance of *on-off goal-directed training*, already successfully employed, e.g., in [20, 21]. Such a method has the drawback of potentially reducing the precision in the prediction of the intended activations due to the delay required by the subject to adapt; nevertheless, it is an accepted way to associate an intended activation with a specific input signal pattern; in the case of amputees, it is actually the only possible way, since amputees cannot produce reliable ground truth in principle.

Participants and experimental protocol

The dataset was gathered from 10 intact subjects induced to follow movements of a realistic 3D hand model displayed on a monitor. The sequence of movements consisted of thumb rotation, flexion of the index and little finger, wrist flexion, extension and supination as shown by Fig. 1a. Each participant repeated this sequence of movements ten times while sitting in front of a monitor (see Fig. 1b).



(a) (b)

Figure 1: (a) Movements executed by the participants. (b) A bird's eye view of the experimental setup used in [5] (reproduced with permission).

EXPERIMENTAL RESULTS

Table 2: Number of participants for which each method performed best per movement.

Metric	Thumb	Index	Little	Wr. flex	Wr. ext.	Wr. sup.
d_{Eucl}	1	4	4	**6**	3	**6**
d_{KL}	**8**	**6**	**5**	2	**4**	3
d_{SPD}	1	-	1	1	3	-
d_{logSPD}	-	-	-	1	-	1
d_{Chordal}	-	-	-	-	-	-

We applied tensor Gaussian Process Regression using RBF kernels based on Kullback-Leibler divergence (d_{KL}), Euclidean metric on $\mathcal{S}_{++}$ (d_{SPD}), log-Euclidean metric on $\mathcal{S}_{++}$ (d_{logSPD}) and Chordal distance (d_{Chordal}) in order to predict the visual stimulus values from the tactile bracelet. We compared the different results with those obtained by applying Ridge Regression (*RR*), equivalent to regularized least squares regression, and Gaussian process with the standard Euclidean metric computed by vectorizing the input data $d_{\text{Eucl}} = \|\text{vec}(\boldsymbol{X}_{(m)}), \text{vec}(\boldsymbol{X}'_{(m)})\|^2$. Cross-validation was applied to obtain a statistically significant estimation. The entire dataset for each participant and movement was randomly shuffled, then 10% of it was used to train each model and the test was performed on the remaining 90%. This procedure was repeated 50 times with a different random shuffle each time.

Table 1 shows three examples of typical average and standard deviation of the normalized root-mean-square error (NRMSE) values obtained by applying the different regression methods for each movement. The NRMSE values for Ridge Regression range from 1% to 11% depending on the movements and participants. This is in line with the values found by Koiva et al. [5]. As expected, all kernel methods achieve better results than Ridge Regression as they can encode nonlinear relationships. The NRMSE values of kernel methods range from 0.5% to 7.5% in function of the movements and participants.

We then compared the efficiency of the different kernel methods for each movement. Table 2 shows the number of participants for which each method performed best per movement. We observe that GP regression using the Euclidean metric and tensor GP regression with KL divergence-based kernel perform the best detection in most of the cases. Tensor GP regression with KL divergence-based kernel is generally the most efficient method to predict finger movement, especially thumb rotation, for most of participants. However, GP with Euclidean-based kernel seems more suitable to detect wrist movements.

DISCUSSION AND CONCLUSION

In this paper, we studied different kernels for tensor Gaussian Processes regression to detect hand and wrist activity by observing muscles bulging in the forearm. The paper concentrated on comparing several regression methods to data obtained in a previous experiment. The results presented above indicate that TMG data obtained from the forearm using the tactile bracelet can be effectively used to obtain *graded* muscle activations — as opposed to classification. As expected, due to the location of involved muscles, NRMSE values for movement involving the wrist are generally better predicted than the fingers movements, and errors in the prediction of thumb rotation are slightly higher. The slightly better results obtained by GP with Euclidean-based kernel on the wrist movements may be explained by the fact that muscles activation are more difficult to measure for finger movements, so that taking the structure of the data in account improves the detection of movements inducing patterns more difficult to distinguish.

All in all, it seems reasonable to claim that such a rich flow of information as the one obtained using 320 tactels (as opposed to the traditional sEMG schema in which a few sensors are involved) can provide better control than the state of the art; particularly, simultaneous and proportional control would greatly benefit from regression applied to TMG data. The results are promising, with all kernels methods being able to predict the different movements with an accuracy superior to previous approaches. Accounting for the structure and geometry of the data proved to be particularly helpful to detect finger movements.

ACKNOWLEDGEMENTS

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Table 1: Normalised root-mean-square error [%] for three participants and each performed movement. The lowest NRMSE for each movement and each participant is highlighted.

Participant 1	Thumb	Index	Little	Wr. flex	Wr. ext.	Wr. sup.
RR	6.75 ± 0.59	7.16 ± 0.55	7.79 ± 0.67	4.54 ± 0.40	4.03 ± 0.58	2.64 ± 0.31
GP, d_{Eucl}	3.79 ± 1.12	2.69 ± 0.43	2.70 ± 0.38	$\mathbf{2.75 \pm 0.67}$	1.86 ± 0.51	$\mathbf{1.19 \pm 0.20}$
TGP, d_{KL}	$\mathbf{2.80 \pm 1.25}$	$\mathbf{1.54 \pm 0.50}$	$\mathbf{1.86 \pm 0.47}$	3.73 ± 1.38	$\mathbf{1.56 \pm 0.50}$	$\mathbf{1.19 \pm 0.46}$
TGP, d_{SPD}	3.66 ± 0.70	2.86 ± 0.50	2.89 ± 0.51	2.94 ± 0.90	1.98 ± 0.52	1.43 ± 0.31
TGP, d_{logSPD}	3.88 ± 0.97	2.94 ± 0.47	2.99 ± 0.54	3.07 ± 0.66	2.09 ± 0.45	1.45 ± 0.29
TGP, d_{Chordal}	3.96 ± 0.87	3.12 ± 0.56	3.10 ± 0.43	2.99 ± 0.73	2.40 ± 0.53	1.51 ± 0.25
Participant 3	**Thumb**	**Index**	**Little**	**Wr. flex**	**Wr. ext.**	**Wr. sup.**
RR	10.82 ± 0.71	9.58 ± 0.81	8.74 ± 0.84	4.61 ± 0.55	3.70 ± 0.26	3.01 ± 0.43
GP, d_{Eucl}	5.12 ± 0.97	$\mathbf{4.66 \pm 0.91}$	$\mathbf{3.98 \pm 0.74}$	$\mathbf{2.80 \pm 0.61}$	1.78 ± 0.22	1.91 ± 0.41
TGP, d_{KL}	7.46 ± 2.13	6.89 ± 1.04	4.75 ± 0.93	3.92 ± 0.86	3.25 ± 0.75	2.71 ± 0.58
TGP, d_{SPD}	$\mathbf{5.07 \pm 0.86}$	5.04 ± 0.92	4.10 ± 0.83	2.83 ± 0.50	$\mathbf{1.68 \pm 0.30}$	1.83 ± 0.47
TGP, d_{logSPD}	5.66 ± 1.13	4.78 ± 0.55	4.27 ± 0.62	3.35 ± 0.56	2.48 ± 0.38	$\mathbf{1.81 \pm 0.33}$
TGP, d_{Chordal}	5.45 ± 0.97	4.75 ± 0.62	4.20 ± 0.68	3.35 ± 0.47	2.75 ± 0.54	1.95 ± 0.39
Participant 6	**Thumb**	**Index**	**Little**	**Wr. flex**	**Wr. ext.**	**Wr. sup.**
RR	4.34 ± 0.25	4.58 ± 0.29	3.15 ± 0.20	1.05 ± 0.06	1.30 ± 0.11	0.98 ± 0.06
GP, d_{Eucl}	1.90 ± 0.35	1.70 ± 0.35	1.23 ± 0.23	0.67 ± 0.24	0.53 ± 0.12	0.45 ± 0.10
TGP, d_{KL}	$\mathbf{1.27 \pm 0.34}$	$\mathbf{1.11 \pm 0.45}$	$\mathbf{0.69 \pm 0.16}$	$\mathbf{0.60 \pm 0.91}$	$\mathbf{0.43 \pm 0.22}$	$\mathbf{0.34 \pm 0.14}$
TGP, d_{SPD}	2.08 ± 0.42	2.06 ± 0.44	1.35 ± 0.21	0.66 ± 0.24	0.48 ± 0.11	0.47 ± 0.08
TGP, d_{logSPD}	2.31 ± 0.33	2.15 ± 0.44	1.75 ± 0.26	0.59 ± 0.16	0.68 ± 0.17	0.65 ± 0.07
TGP, d_{Chordal}	2.13 ± 0.32	2.12 ± 0.34	1.67 ± 0.21	0.61 ± 0.09	0.77 ± 0.16	0.62 ± 0.078

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IMPLANTED MAGNETS TRACKING AS A NOVEL METHOD FOR PROSTHETIC HANDS CONTROL

Sergio Tarantino, Francesco Clemente, Diego Barone, Marco Controzzi, and Christian Cipriani
The BioRobotics Institute, Scuola Superiore Sant'Anna, V.le R. Piaggio 34 56025 Pontedera, Italy

ABSTRACT

Restoring dexterous motor functions equivalent to that of the human hand after amputation requires the implementation of an effortless Human-Machine Interface that bridges the artificial hand to the sources of volition. New research approaches span from invasive interfaces (e.g. peripheral nerve electrodes) to techniques aimed at increasing the number of independent signal sources available for control (e.g. targeted muscle reinnervation). Among those, solutions based on Implantable Myoelectric Sensors (IMES) are very promising. IMES are small electrodes that wirelessly transmit intramuscular EMG signals to the prosthesis. Their main drawback is that they need to be powered wirelessly with unavoidable power consumption. As an alternative solution, within the framework of the MYKI project (ERC-SG #679820), we propose to implant small magnetic markers (MMs) directly in forearm muscles. By doing so, it is possible to detect the muscles deformation during contraction by tracking the MMs using a localizer and use such information to drive a prosthetic limb. We dubbed this the MYoKInetic (MYKI) interface (Fig. 1).

In order to test the feasibility of the MYKI interface, we built a physical mock-up (PMU) of the human forearm aimed at reproducing its muscles' natural position and deformation. Muscles were modelled as a wire attached on one side to a servo motor. Four MMs were attached to four wires in the PMU, associated to four muscles actuating the adduction/flexion of the thumb and flexion of index and middle fingers. The localizer comprised two printed circuit boards, each equipped with three 3-axis magnetic field sensors (HMC5983, Honeywell International Inc.) and a host PC used to solve the magnetic inverse problem (i.e. retrieving the 3D position of the four MMs using the magnetic sensors readout) at 25 Hz. The maximum accuracy and repeatability errors of the system were found to be 16% and 1% the mean stroke of the MMs (13mm), respectively. Albeit large, the accuracy is of minor importance for this system as it relies on the ability to discriminate a movement. The geomagnetic field also affected the system reliability (100% error). However, simulations showed that such disturbances can be attenuated by using a magnetic shield (μ_r=200000, residual error 0.5%). Future work will focus on increasing the number of MMs tracked and study the influence of magnets orientation.

Implantable MMs are not subject to failure and don't need to be powered, potentially increasing the life time of the implant with respect to previous solutions.

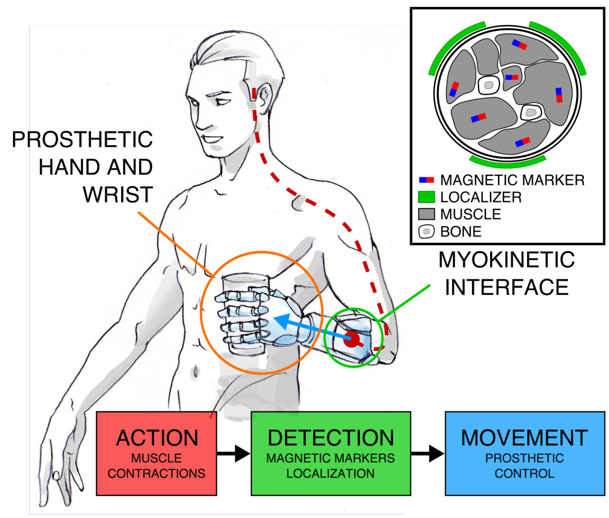

Figure 1 Overview of the MYoKInetic interface.

EVALUATION OF HAND FUNCTION TRASPORTING FRAGILE OBJECTS: THE VIRTUAL EGGS TEST

Marco Controzzi, Francesco Clemente, Neri Pierotti, Michele Bacchereti and Christian Cipriani

The BioRobotics Institute, Scuola Superiore Sant'Anna, V.le R. Piaggio 34 56025 Pontedera, Italy

ABSTRACT

The evaluation of the hand function is of great importance in both clinical practice and research activities. Assessment tools are essential to provide the therapist or investigator with relevant and objective information concerning the patient status, the effectiveness of the treatment program and the assistive technology prescribed. This abstract presents the design and the administrative instructions of a new hand assessment test: the *Virtual Eggs Test* (VET), that resembles the task of transporting fragile objects. The test builds on investigations on pick and lift tasks, showing that humans exert on objects grip forces (GF) that are sufficient to prevent slips, and yet are not so excessive as to crush a fragile object. While grasping humans apply GF and load forces (LF) in coordination, which is disrupted when sensory information from the fingertips is lost. The VET replicates the box and blocks test except that breakable blocks are used instead of the standard wooden ones (Figure 1a). The performance is measured by the number of blocks transferred and percentage of blocks broken during 1 minute trials.

We designed two assessment instruments for this test that may be used depending on the performance that have to be recorded: the *magnetic Virtual Egg* (mVE), and the *instrumented Virtual Egg* (iVE). In the mVE (Figure 1b), empty blocks (40x40x40mm, ~80g) are equipped with a magnetic fuse which exploits the attraction force between two magnets to maintain a fixed distance between two opposite walls of the block. When a GF larger than the attraction force between the two magnets is exerted on the object, the walls collapse and the object "breaks". The iVE (Figure 1c) enriches the assessment power of the mVE by measuring the GF and LF. This allows to evaluate (i) the ability to modulate the GF and (ii) the rate of temporal GF-LF coordination. The iVE is a test-object (57×57×57mm; variable weight from ~180g up to 340g) equipped with two strain gauge-based force sensors, able to measure the GF. An additional sensor is placed on the base of the test object, acting as a stand able to measure the LF of the test-object when it is resting on it. If the subject generates a GF larger than the threshold, the instrumented object virtually breaks; this may be signalled to the subject through an acoustic signal, coloured light and/or short vibration. Data are recorded and transferred using a wireless protocol to the PC.

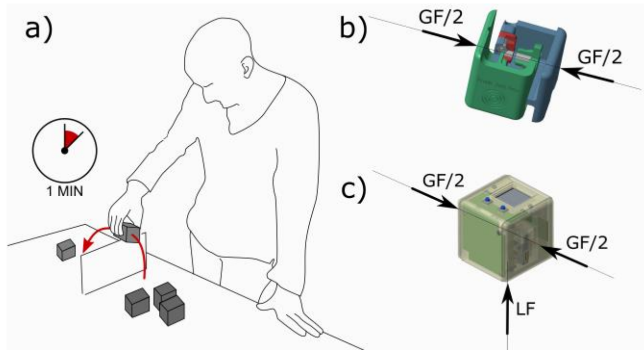

Figure 1- a) subject performing the VET, b) the *magnetic Virtual Egg* (mVE), c) the *instrumented Virtual Egg* (iVE).

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THE POINT DIGIT: A PASSIVE, RATCHETING PROSTHETIC FINGER MANUFACTURED USING METAL LASER SINTERING TECHNOLOGY

[1]Jacob Segil, [2]Stephen Huddle, [2]Levin Sliker and [2]Richard Weir

[1]University of Colorado at Boulder
[2]University of Colorado at Denver

ABSTRACT

Partial hand amputees represent the largest population of individuals with upper-limb amputation. However, commercially available prosthetic finger devices lack strength, durability, and anatomically correct centers of rotation. We developed a passive, ratcheting prosthetic finger that is manufactured using metal laser sintering rapid manufacturing technology, the Point Digit.

The Point Digit is a purely mechanical prosthetic finger (i.e., no actuation, electronics, etc.). A ratchet positions the finger into one of ten distinct levels of flexion. The ratchet ensures that the finger is non-backdrivable and can withstand 250 lbs. (1.1 kN) of static load without failure. An EOS M270 direct metal laser sintering 3D printer manufactures most components of the Point Digit using maraging steel powder (EOS MaragingSteel MS1). This material has a yield strength of 152 ksi and modulus of elasticity of 23 Msi ensuring ample mechanical strength. An internal honeycomb structure maintains strength and reduces weight of the individual components. The average finger mass for all lengths of the Point Digit is 45g fully assembled. The length of the Point Digit can be scaled between 80 mm to 100 mm in length, which nearly encompasses the ranges of finger lengths for 25 – 75th percentile males and 50-100th percentile females.

The extension of the Point Digit can occur in two ways: 1) a self-locking button releases the ratchet and extends the finger or 2) the full flexion of the finger causes the finger to spring-back to full extension. By allowing external features to position the finger in both flexion and extension, the contralateral limb is not required for use of the Point Digit.

A kinematic linkage system couples all three joints of the Point Digit. This linkage system flexes all three joints at an anatomically appropriate rate. The linkage system ensures a finger that behaves similar to the intact limb while maintaining mechanical strength.

Finally, the Point Digit provides an anatomically appropriate center of rotation of the metacarpophalangeal joint. This feature ensures a clinically sound system that easily integrates into a prosthetic socket. A mounting bracket provides a method for prosthetists to install the Point Digit into the prosthetic socket and ensure appropriate positioning of the finger with respect to the physiological limb.

The Point Digit advances the field of prosthetic finger design by providing a durable, ratcheting prosthetic finger using metal laser sintering technology.

SHROOM TUMBLER GROUND REACTION FORCES

E. Biden, W. Hill, G. Bush, J. Porter, V. Chester
Institute of Biomedical Engineering, and McCain Human Performance Laboratory
University of New Brunswick
Fredericton, NB, Canada

ABSTRACT

Our clinic has two girls with congenital right transradial limb loss who use the Shroom Tumbler from TRS as their terminal device for competitive cheerleading. The younger of the two had complaints of discomfort in her arm when doing particular tumbling activities. Our objective was to understand what the forces were which were carried by the Shroom as compared to the forces in their sound hands.

The literature is sparse for hand to floor contact forces and we have found no reports for amputees. We have found that the range of observed loads in the hands and wrists of normally limed individuals doing things like back handsprings ranges from less than body weight to about four times body weight.

We used a 12-camera Vicon motion capture system with Kistler force plates to measure various floor exercises used in cheerleading, such as cartwheels and handsprings.

Initial tests showed high forces during tumbling activities. Subsequent changes to the prosthesis included a modified socket which improved her comfort level so that she no longer complains of pain in her sound limb and her prosthetic side. The socket, liner and distal end pads were customized to provide a total contact hydrostatic socket fit. This type of strategy provides a limb, socket interface that can withstand the high peaks in ground reaction forces and provides improved proprioception and stability for gymnastic/cheer activities.

The Shroom provides a good terminal device for cheerleading floor activities but using it should be coupled with coaching which understands the mechanics of the moves and proper form coupled with prosthetic care which can provide an optimal fitting.

INTRODUCTION

Our clinic has two girls with congenital right transradial limb loss who use the Shroom Tumbler from TRS as their terminal device for competitive cheerleading. The younger of the two had complaints of discomfort in her arm when doing particular tumbling activities during cheerleading. Our objective in testing them was to understand what the forces were which were carried by the Shroom as compared to the forces in their sound hands.

The literature is relatively sparse for hand to floor contact forces and we have found no reports for amputees. Penitente and Sands report some upper limb results and suggest that loads on the hands and arms should be considered "high" between 0.86 and 1.81 body weight (BW) for movements such as back handsprings. This is consistent with Davidson et al who report impact forces between 1.6 and 2.4 BW but are lower than Burt et al who report loads of up to 4 time BW in National and International level gymnasts. Thus this clinical study is, we believe, the first to measure ground reaction forces in upper limb amputees.

CLINICAL PROBLEM

The Shroom is very compliant when contact is made on the edge but is much stiffer if it is loaded axially along the arm.

In these types of prosthetic fittings the limb is axially loaded and subjected to high loading forces. Usually we think of an upper limb prosthesis as being subjected to distraction forces. However, in these fittings the upper limb fitting is treated more like the fitting of a lower limb prosthesis.

Tumbling activities happen very quickly and are hard to observe without motion and force capture technology. We have access to a 12 camera VICON motion capture system with force plates which was used to track the motion and record the forces through the hands and arms. Ages, height and weight are shown in Table 1.

Table 1: Information about the participants.

Participant	Age	Height (cm/in)	Weight (Kg/#)	Prosthesis
CD	7	124.5/ 49"	216N/48#	Right
EB	10	145/ 57"	477N/107#	Right
SH	11	144/ 58.5"	460N/103#	Nrm Limbs

The participants performed a variety of moves as listed in Table 2. CD is younger and at an earlier stage in learning moves and so had some moves which she did not perform. EB had injured her sound arm earlier in the year and was not comfortable doing Back Handsprings.

Table 2: Activities performed

Participant	Cart Wheels	Back Walkover	Front Walkover	Back Handsprings	Handstands
CD	Y	Y		Y	
EB	Y	Y	Y		Y
SH	Y	Y	Y	Y	Y

The measurements taken were primarily the forces recorded by the force plates for each of the participants.

We were concerned that for back handsprings and cartwheels, the forces for the younger girl were substantially higher than body weight, whereas the normally limbed participant and the older Shroom user kept their forces well below body weight.

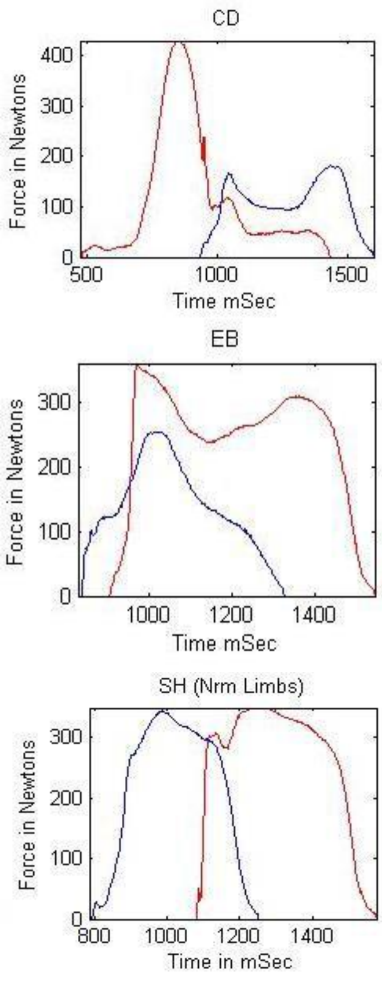

Figure 1: Cartwheels for the two Shroom users CD and EB and the normally limbed SH.

Back handsprings by the normally limbed individual and the younger Shroom user showed substantial differences. The normally limbed girl was very consistent and symmetrical with a maximum initial contact force more than double body weight (BW). The younger Shroom user experienced 4.45 times BW on the sound side and 2.3 times BW on the prosthesis. There were very high loading rates for the prosthesis. The younger Shroom user's loading rates during back handsprings were as high as the normally limbed girl who weighs twice as much.

We had an experienced Cheerleader/tumbling Coach examine the data and videos of the testing session. She recommended changes in the training program and the use of more active spotting during potentially high force maneuvers such as back handsprings.

The prosthetist modified the socket used with the Shroom by making a custom silicone distal end pad. This child has an Ohio Willow Wood Spirit liner which is covered with a fuzzy pile fabric similar to Velcro. The inside of the socket had a patch of hook Velcro inserted which prevented the socket from rotating during vigorous Cheer activities.

The younger participant was seen again in October for a follow-up which provided the opportunity for a retest and another session with the cheer coach.

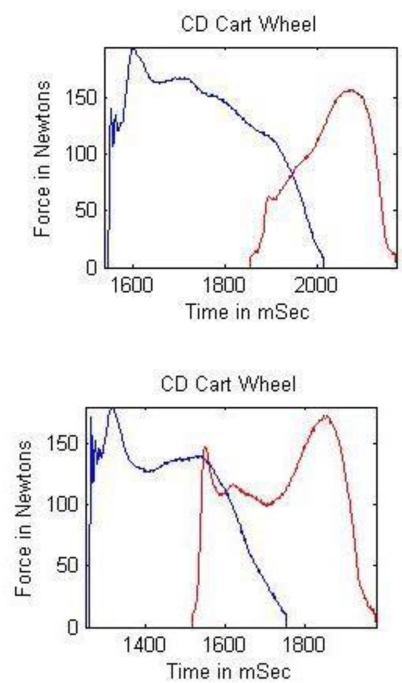

Figure 2: Typical cartwheels for CD during October visit.

As can be seen in the examples of cartwheels in figure 2, CD is no longer having very high loading during the activity. The sessions with the cheer coach have improved her arm positioning when doing many of the floor activities.

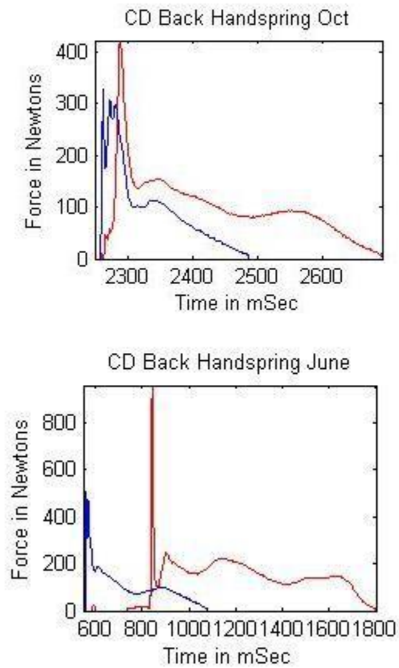

Figure 3: Back Handsprings in the June data collection at the top and for October on the bottom.

Note the very high June spikes. For October it is important to note that this was with spotting which would have reduced the forces to some degree.

In Figure 3 the other major change is that in June the prosthesis was landing well in advance of the sound hand while in October they were landing together which reduces the load on each side.

DISCUSSION AND CONCLUSION

Improvements to the prosthesis include the modified socket which has improved her comfort level so that she no longer complains of pain in her sound limb and her prosthetic side. The socket and liner and distal end pads were customized to provide a total contact hydrostatic socket fit. This type of strategy provides a limb socket interface that can withstand the high peaks in GRFs and provide improved proprioception and stability for gymnastic/cheer activities.

We have observed that the participants are not as consistent as we would have thought before doing the tests. There is a substantial amount of variation from move to move. The second set of cart wheels, however, never reaches the levels which were seen, and caused concern, initially.

The TRS Shroom provides a good terminal device for cheerleading floor activities but using it should be coupled with coaching which understands the mechanics of the moves and proper form coupled with prosthetic care which can provide an optimal fitting for the activities.

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DISCLAIMER

This study used clinical standards of care, and was not a research study. All results and pictures report only de-identified data, unless informed consent was obtained.

This secondary analysis of data has been reviewed by the Research Ethics Board of the University of New Brunswick and is on file as REB 2017-059.

KINEMATIC INSIGHTS FROM A NOVEL GAZE AND MOVEMENT METRIC FOR UPPER LIMB FUNCTION: NORMATIVE AND PROSTHETIC COMPARISON

Aida Valevicius, Quinn Boser, Ewen Lavoie, Craig Chapman, Patrick M. Pilarski, Albert Vette and Jacqueline Hebert

University of Alberta

ABSTRACT

The evaluation of advanced upper limb prosthetic devices is limited since current outcome metrics may not be sensitive enough to detect compensatory movements and control strategies. We developed two functional tasks for device and performance assessment that are amenable to motion and eye tracking that mimic activities of daily living. The tasks incorporate elements of lateral motion, crossing the body's midline, accuracy, and risk.

Kinematic data from twenty healthy participants and one prosthetic user with a myoelectric and body-powered prosthesis were analyzed. The following degrees of freedom were included in the analysis: trunk flexion-extension, abduction-adduction, and axial rotation; shoulder flexion-extension, adduction-abduction, and internal-external rotation; elbow flexion-extension and pronation-supination; and wrist flexion-extension and ulnar-radial deviation. The range of motion (ROM) was extracted from joint angle trajectories for each examined degree of freedom. End-effector metrics included time to task completion, maximum hand velocity, time and percent to peak velocity, and number of movement units.

Joint kinematics and end-effector metrics where substantially different between normative and prosthetic performance. In comparison to normative performance, the prosthetic user exhibited increased ROM in trunk flexion-extension and shoulder adduction-abduction for both prosthetic devices and tasks. This suggests a compensation for lack of elbow and wrist joint motion by relying more on trunk and shoulder motion to complete the tasks successfully. Despite longer movement times, when using his myoelectric prosthesis, the prosthetic user's ROM was closer to normative performance when compared to his body powered prosthesis. Velocity peaks occurred earlier during reach and grasp movements, indicating a prolonged deceleration phase or a change in movement strategy.

These preliminary results suggest that a range of quantitative information can be extracted from a kinematic analysis of upper body movements. Movement strategies that trend towards normative functional motion could have the potential to reduce the risks of overuse injuries in prosthetic users, given that repetitive movement outside of the normal ranges of function put individuals at greater risk (Kidd, McCoy, & Steenbergen, 2000). In fact, Jones & Davidson reported that 50% of individuals with upper limb amputations reported suffering from an overuse injury (Jones & Davidson, 1999). The proposed motion capture protocol allows us to assess differences between normative and prosthetic performance, but also between different prosthetic technologies. Like common gait assessment practices, the norms can be used as a benchmark for assessing upper limb impairments, advanced technologies, and performance improvements over time, which will be the focus of future work.

TWO-DOF, DYNAMIC EMG-BASED ESTIMATION OF HAND-WRIST FORCES WITH A MINIMUM NUMBER OF ELECTRODES

Chenyun Dai[1], Ziling Zhu[2], Carlos Martinez-Luna[3], Thane R. Hunt[3], Todd R. Farrell[3], Edward A. Clancy[2]

[1]*University of North Carolina-Chapel Hill and North Carolina State University, USA*
[2]*Worcester Polytechnic Institute, Worcester, MA, USA*
[3]*Liberating Technologies, Inc., Holliston, MA, USA*

ABSTRACT

Introduction:
Commercial hand-wrist prostheses realize partial function for amputees via electromyogram (EMG) control derived from remnant muscles. Most EMG-based prostheses provide only one degree of freedom (DoF) of control at a time. Recent studies have used different approaches to overcome this challenge, with two main limitations. First, most studies used a large number of high-density electrodes. None of them studied the possibility of minimizing the number of electrodes for practical commercial prosthesis use. Second, very few studies have investigated the feasibility of 2-DoF control for the hand and wrist concurrently. Our study explored the minimum number of electrodes required for 2-DoF simultaneous hand-wrist force estimation.

Methods:
Nine able-bodied subjects participated. Sixteen conventional bipolar EMG electrodes were equally spaced around the proximal forearm. The subject's hand was secured to a load cell which measured hand open-close (Opn-Cls) force, and their wrist was fixed to a 3-DoF load cell which measured extension-flexion (Ext-Flx), radial-ulnar deviation (Rad-Uln) or pronation-supination (Pro-Sup) force. The subject was required to perform constant-posture, dynamic force tracking based on a computer-generated random moving target (0.75 Hz bandwidth). First, 1-DoF trials tested the four forces separately. Second, 2-DoF trials tested hand Opn-Cls always paired with one of the three wrist forces. Each task had four trials, two of which were used for training and two for testing, of a linear least squares regression EMG-force model. Backward stepwise selection was used to reduce the number of electrodes from 16 to 1.

Results:
For the 1-DoF models, two-way RANOVA found an effect due to number of electrodes [F(1.8, 14.7) = 99, p_{GG} <0.001], but not DoF [F(3, 24) = 0.54, *p*=0.66]. *Post hoc* paired *t*-tests (Bonferroni corrected) only found error higher when comparing 1 electrode to more than 1 ($p \leq 0.001$); and 13 electrodes to 10 (*p*=0.006; this difference is argued to be a false positive). The errors for the four respective forces, Opn-Cls, Ext-Flx, Rad-Uln and Pro-Sup, were 8.8 ± 3.3, 8.3 ± 2.0, 9.0 ± 1.6 and 8.7 ± 2.2 %MVC.

For 2-DoF models trained from 1- and 2-DoF trials and tested on 2-DoF trials, the RANOVA main effect of number of electrodes was significant [F(1.6, 12.9) = 99, $p_{GG} = 10^{-6}$], but DoF was not [F(2, 16) = 0.07, *p*=0.9]. *Post hoc* analysis of number of electrodes only found that 1 electrode exhibited higher error than more than 1 (*p*<0.003), 2 electrodes higher than more than 3 (*p*<0.003), 3 electrodes higher than more than 5 (*p*<0.02), 4 electrodes higher than more than 5 (*p*<0.03), and 5 electrodes higher than 6 or 10–13 (*p*<0.05). With four electrodes, the 2-DoF errors for Ext-Flx, Rad-Uln and Pro-Sup (each paired with hand Opn-Cls) were 9.2 ± 2.0, 9.2 ± 1.6 and 9.2 ± 1.4 %MVC, respectively.

Conclusion:
While low errors in a lab study do not necessarily reflect improved performance in a prosthesis, such studies in able-bodied subjects are useful in refining algorithms before undertaking more expensive field studies using a prosthesis. Our 2-DoF results showed a similar error level as our 1-DoF results. As few as four conventional electrodes provided good performance for estimating 2-DoF simultaneous hand-wrist forces. Testing these techniques in a hand-wrist prosthesis is an appropriate next research step.

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PATTERN RECOGNITION CONTROL OF THE DEKA ARM IN TWO TRANSHUMERAL AMPUTEES WITH TARGETED MUSCLE REINNERVATION

Linda Resnik, [1,2] Jill Cancio, [3,4] Christopher Fantini, [5] Andrea Ikeda,[3,4] Nicole Sasson, [6, 7]
1 Providence VA Medical Center, 2 Brown University, 3 Center for the Intrepid, Brooke Army Medical Center, [4]Extremity Trauma and Amputation Center of Excellence (EACE), [5] JJ Peters VA Medical Center, [6] VA NY Health Harbor System, [7] NYU School of Medicine

ABSTRACT

Background: Recent utilization of EMG pattern recognition (PR) as a control input for the DEKA Arm shows promise for decreasing cognitive burden by eliminating the foot controls. **Purpose:** To report outcomes and experiences of two subjects with transhumeral (TH) amputation who had undergone targeted muscle reinnervation and were fit with, and trained to use, a DEKA Arm, with 5 degrees of freedom (DOF) controlled by EMG PR. **Methods:** This study had 2 portions: in-laboratory training (Part A) and home use (Part B). Quantitative outcomes and qualitative data were collected at baseline, end of Part A and end of Part B. **Results:** Both subjects controlled a 5 DOF DEKA Arm using EMG PR and were generally satisfied with this control method. Quantitative outcomes were mixed. Subjects provided feedback on the DEKA Arm. **Conclusion:** PR control for the DEKA Arm was feasible in persons with transhumeral amputation who have undergone TMR surgery given adequate training.

INTRODUCTION

The DEKA Arm was designed to utilize unique “strap and go” controls including inertial measurement units (IMUs) worn on the feet and pneumatic pressure transducers. [1] To date, one or more IMUs were required to operate the DEKA Arm. Our prior study found that there was substantial cognitive load required for the complex task of pre-planning and sequentially controlling multiple joints/actuators using these controls. Recent commercial release of an EMG pattern recognition (PR) system, the Coapt System, and adaptation of that system for control of the DEKA Arm, holds promise for decreasing cognitive burden by eliminating the IMUs.

Substituting foot control with more intuitive PR control may lead to greater acceptance of, and better function with, the DEKA Arm. However, the benefits need to be examined empirically. This paper describes outcomes and experiences of 2 subjects with transhumeral (TH) amputation who had undergone targeted muscle reinnervation (TMR) who used a DEKA Arm with 5 degrees of freedom (DOF) controlled by EMG PR: hand open/close, wrist flexion/extension, pronation /supination, elbow flexion/extension, humeral internal/external rotation.

METHODS

Study Design

The study was approved by the Institutional Review Boards at participating sites. The study consisted of 2 portions: in-laboratory training (Part A) and home use (Part B). Subjects were tested with their personal prosthesis (if applicable) at baseline and with the DEKA Arm at the end Part A and at monthly intervals during Part B.

Subjects

Both subjects underwent TMR surgery approximately 5 years prior. Subject 1 had previously participated in the VA Study to Optimize the DEKA Arm and had been trained to use the DEKA Arm using IMU controls. At baseline, he was no longer using a prosthesis. Subject 2 was tested with his 3 DOF myoelectric prosthesis which he controlled using PR via standard Coapt controls.

Controls Set-up and Prosthetic Training

The Coapt system was configured to communicate directly with the DEKA Arm by way of bi-directional digital bus. Subjects were fit and trained to use the humeral configuration of the DEKA Arm using these controls. By the end of Part B, both controlled 5 DOFs (humeral rotation, elbow flexion/extension, pronation/supination, wrist flexion/extension and hand open/closed) with PR. Both controlled grip selection by using pressure transducer. During Part A Subject 1 had EMG PR control of humeral rotation however, it was controlled using the same pattern that he used for wrist flexion/extension pattern by switching from hand to arm mode using a pneumatic pressure transducer. His controls where updated requiring a unique muscular recruitment pattern for humeral rotation and eliminating the need for mode switching, when the new prototype became available at the beginning of Part B.

Subject 1, had 19 hours training with an early prototype of the Coapt controls, and 4 additional days of training with the new prototype. He then participated in approximately 12 weeks of home study. However, he decided not to wear the DEKA Arm after the 4th week of the home study (see results below for reasons why). Subject 2 had approximately 34 hours of training with the new Coapt controls, and participated in 12 weeks of home study.

Data Collection

Both qualitative and quantitative data were collected. Qualitative data included recorded audio and video from in-laboratory and testing sessions, as well as data from open-ended surveys and transcripts from semi-guided interviews administered at the end of Parts A and B.

Self-report and performance based measures of function, and measures of quality of life (QOL), prosthesis satisfaction and community integration were collected at baseline (prior to training), at the end of Part A, and at the end of Part B. Self-report measures included a modified version of the Upper-Extremity Functional Scale (UEFS) [2, 3], Patient-Specific Functional Scale (PSFS) [4], the QuickDASH [5], Trinity Amputation and Prosthesis Experience Satisfaction Scale (TAPES) [6], QOL scale [7], and the Community Reintegration of Service Members Computer Adaptive Test (CRIS-CAT) [8]. Performance based measures included the modified Jebsen-Taylor Hand Function Test (JTHF) [3], the Activities Measure for Upper-Limb Amputees (AM-ULA) [9], and the University of New Brunswick Test of Prosthetic Function for Unilateral Amputees (UNB) [10]. Interpretation of measure scores is shown in Table 1.

Figure 1 Subjects wearing DEKA Arm

a. Subject 1 b: Subject 2

Data Analysis

We gathered key impressions about: PR control, DEKA Arm, and comparisons to the existing prosthesis (when applicable) to generate a detailed report of subjects' perspective. Highlights were selected to illustrate major findings. Descriptive analyses of outcomes were conducted.

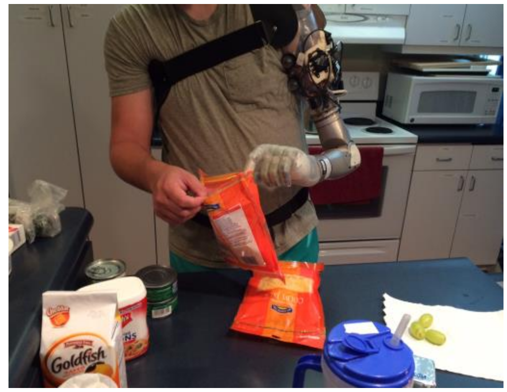

Figure 2: Subject 1 in training with check socket

Figure 3: Subject 2 using DEKA Arm

RESULTS

Subject 1

This subject was the first to utilize an updated version of the Coapt system to control the DEKA Arm. During the first few weeks at home, he experienced several technical problems related to the new Coapt controls configuration which required him to return the device for repair and also travel to the site to adjust the configuration and perform additional training with a new set-up which, as described above, required him to use a unique muscular pattern to control humeral rotation. The subject never fully acclimated to this change, and stated that this change, "made me not want to use the thing at all." The subject stated that the number of movements he needed to control was "extensive" and adding one additional movement "changed everything." That said, at the end of the study the subject indicated that he was "very happy" with the Coapt controls, but felt that he would have benefited from additional training.

He also compared PR control of the DEKA Arm to IMU control (which he had used previously) very favourably saying that it was, "so much more intuitive", and that "*it beat out the IMUs a billion to one"*. This subject reported that he disliked the weight of the DEKA Arm, which he called "excessively heavy."

This subject disliked the external battery which made donning and doffing more cumbersome and made wearing it "*so constricting*." He also complained about the bulk of the device and the excessive external wiring, which he reported made the device look "*horrible*." He also reported that if he was actively using the device and he was sweating he might have to doff and re-don it because he would lose function due to poor electrode contact. Finally, he felt that the battery life was insufficient and with the EMG PR system use , the battery was drained within a couple of hours.

Subject 1 rated his satisfaction with the DEKA Arm as 3.1, neither satisfied or dissatisfied), on a scale of 1-5 at the end of Part A (when he was using a wrist flexion/extension pattern to control humeral rotation) and as 1.9 (dissatisfied) at the end of Part B (when using a humeral rotation pattern to control humeral rotation). Self-reported difficulty performing activities, as measured by the PSFS was decreased at the end of Part B as compared to Baseline. Perceived difficulty with functioning (UEFS) was improved at the end of Part B, however perceived disability (QuickDASH) was unchanged from baseline. Community integration was improved at the end of Part B as compared to baseline and end of Part A. QOL was similar across time. (Table 1). Dexterity, as measured by the modified Jebsen Taylor Test, improved on 4 subtests at the end of Part B as compared to end of Part A. Activity performance, as graded by the AM-ULA, was improved. (Table 2).

Subject 2

Subject 2 rated satisfaction with the DEKA Arm similarly to the way he rated his own prosthesis at baseline (satisfied). His comments indicated that he found that the DEKA Arm was heavy and wished that the internal battery life was longer. Although he was an experienced PR user, he stated that there was "a steep learning curve" required to control the additional degrees of freedom of the DEKA Arm (humeral rotation and wrist flexion/extension). However, at the end of the study he felt that he had acclimated "*completely*" to the controls, and he rated his skill level as "good", although he commented that he found that the device was more "*mentally taxing*" than his own device. He participated in the study during hot summer months and reported that sweating impacted the precision of his controls, "*..small movements become real erratic and large, so I found like the Arm was moving unintentionally….. so I would constantly have to correct."* He reported that he experienced the same heat-related control issues with his personal prosthesis, and that he typically did not wear his myoelectric in summer months because it was too hot and he was very active. Subject 2 stated that it was easier to perform daily activities with the DEKA Arm given its wrist flexion and humeral rotation, which he said made the prosthesis "look and feel more natural." These additional degrees of freedom, he felt, allowed him to reduce compensatory movements, "*you don't have to turn your whole body.... you don't have to angle yourself because you have wrist flexion."*

Self-reported difficulty performing activities, as measured by the PSFS, the UEFS, and the QuickDASH, was improved at the end of Part B as compared to baseline. QOL and community integration was similar across time (Table 1). Dexterity, as measured by the JTHF, improved in 2 subtests at the end of Part B as compared to baseline, but was worse in 5 subtests. Activity performance (AM-ULA) decreased from baseline to end of B, but prosthetic skill and spontaneity (UNB) improved slightly (Table 2).

DISCUSSION

This case series describes the experiences and outcomes of two transhumeral amputees who had undergone TMR surgery 5 years prior. Both learned to control a 5 DOF prosthesis (the DEKA Arm) using EMG PR and were generally satisfied with this control method. Subject 1, who had used the DEKA Arm with IMU foot controls clearly preferred EMG PR to the foot controls.

We do not know how user perspectives on the EMG PR controls would have changed with additional training or home use time. Subject 1 was initially trained with an earlier prototype of EMG PR. Although he received additional training, when his control system was updated, he never fully acclimated to this change, and did not use the DEKA Arm regularly at home. Although Subject 2 felt that he had fully acclimated to the DEKA EMG PR controls, he reported that using them was more mentally taxing than using PR for his own, less complex device.

Given that Subject 1 was not a prosthesis user we were unable to compare performance outcomes of the DEKA Arm to his own prosthesis. While we did observe an improvement in his CRIS-CAT scores at the end of Part B, we attribute this to an improved living situation associated with a move. For Subject 2 there were clear improvements in perceived disability using the DEKA Arm. Subject 2's comment that controlling the DEKA arm was more "mentally taxing" than his own prosthesis may be due to the fact that there are more DOF requiring control in the DEKA arm as compared to his existing prosthesis (5 and 3 respectively).

Table 1: Interpretation of Measures

Self-report measures	Interpretation
Patient-Specific Functional Scale (PSFS)	Higher scores indicates less difficulty
Modified Upper-Extremity Functional Scale (UEFS)	Lower scores indicates less difficulty
Upper-Extremity Functional Scale (Use)	Higher scores indicates more activities done with prosthesis
Disabilities of the Arm, Shoulder and Hand Score (QuickDASH)	Higher scores indicate greater disability
Community Reintegration Computer Adaptive test (CRIS-CAT)	Higher scores indicates better community integration
Quality of Life (QOL)	Lower scores indicate worse QOL
Trinity Amputation and Prosthesis Experience Scales (TAPES)	Higher scores indicate greater satisfaction
Performance Measures	
Jebsen-Taylor Hand Function Test (JTHF)	Higher scores indicate better performance
University of New Brunswick Test of Prosthetic Function (UNB)	Higher scores indicate better performance
Activities Measure for Upper-Limb Amputees (AM-ULA)	Higher scores indicate better performance

Both subjects expressed concerns about the weight of the DEKA Arm, short internal battery life, and dissatisfaction with other features such as external cables which clearly tempered their enthusiasm for the DEKA Arm.

Table 2. Outcomes Across Time

	Subject 1			***Subject 2***		
	Baseline	End of A	End of B	Baseline	End of A	End of B
Self-report measures						
PSFS	9.0	6.8	6.6	3.0	7.5	7.0
UEFS	45.4	48.2	33.1	39.1	40.7	32.0
UEF use	0.0	1.0	0.0	0.4	0.4	0.4
QuickDASH	29.5	25.0	29.5	27.3	31.8	20.5
CRIS-CAT						
Extent	33.0	33.0	40.0	58.0	58.0	55.0
Perceived Limitations	39.0	42.0	46.0	55.0	57.0	55.0
Satisfaction	36.0	41.0	49.0	55.0	55.0	55.0
QOL Scale	3.9	4.1	4.1	6.3	6.0	6.2
TAPES Satisfaction	-	3.1	1.9	4.4	3.9	4.4
Performance measures						
JTHF						
Writing	-	0.26	0.32	0.31	0.40	0.53
Page Turning	-	0.05	0.05	0.11	0.08	0.07
Small items	-	0.05	0.15	0.13	0.03	0.03
Feeding	-	0.06	0.09	0.07	0.11	0.09
Light Cans	-	0.09	0.16	0.23	0.07	0.11
Heavy Cans	-	0.18	0.13	0.25	0.09	0.14
UNB:						
Spontaneity	-	3.5	3.3	3.3	3.4	3.7
Skill	-	3.2	3.2	2.9	3.2	3.4
AM-ULA	-	16.7	19.4	22.7	22.7	18.8

CONCLUSION

This case series demonstrated that the adapted PR control system for the DEKA Arm which we developed for this study was feasible for use in persons with transhumeral amputation who have undergone TMR surgery. Findings also suggest that the DEKA Arm may be suitable for this patient population, given adequate training. However the device may be better accepted if it were lighter, had fewer external cables and wires and a longer internal battery life.

ACKNOWLEDGEMENTS

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A NATIONAL STUDY OF VETERANS AND SERVICE MEMBERS WITH UPPER LIMB AMPUTATION: SURVEY DEVELOPMENT AND PILOT TESTING

Linda Resnik[1,2], Sarah Ekerholm[1], Melissa Clark[3]

[1]*Providence VA Medical Center, Brown University*

[2]*Brown University*

[3]*University of Massachusetts Medical School*

BACKGROUND AND PURPOSE

Quality gaps in care to persons with upper limb amputation have been reported. Studies showing dissatisfaction amongst combat Veterans with upper limb loss led to calls for studies to understand needs and improve satisfaction. In 2016, a new longitudinal study was funded to address this gap. The new study includes both telephone surveys and in-person data collection. The purposes of this presentation are to describe the pilot work conducted to test and refine the survey for the new study and report preliminary results.

DESIGN AND METHODOLOGICAL PROCEDURES USED

The pilot study had two phases: survey development/cognitive testing to identify problematic items (Phase 1), and pilot testing of the full survey (Phase 2). The full survey was designed to assess demographics, amputation history, prosthesis use, function, quality of life, satisfaction with prosthesis and amputation care, quality of care, and included a risk-benefit assessment of technological advances requiring surgical intervention. The survey included new items, validated standardized measures, and items modified from a prior study. Cognitive testing and pilot testing resulted in refinements to the survey and a decision to administer by telephone only.

RESULTS

Phase 1 included 10 participants; 90% male, mean age 56 years, 30% with transradial (TR), 60% with transhumeral (TH), and 10% with shoulder level amputation (shoulder); 60 % were prosthesis users. Phase 2 included 13 participants; mean age 59 years, 92% male, 38% TR, 46% TH, and 15% shoulder; 77% were prosthesis users. Amongst Phase II prosthesis users, 60% used a body-powered and 40% a myoelectric/hybrid. Seventy percent used two or more types of devices, and 60% used two or more types of terminal devices. Prosthesis users averaged 3.4 hours of use per day, and 40% were dissatisfied with their prostheses. Twenty three percent indicated that they would be willing to consider surgery for osseointegration, 54% for greater prosthesis control, and 31% for sensory restoration.

CONCLUSIONS/IMPLICATIONS

The refined survey is ready for use. Preliminary findings suggest that despite availability of multiple types of devices, there was a high prevalence of dissatisfaction with devices. At least half of pilot participants indicated their willingness to incur risk to obtain the benefits of a new or emerging prosthetic technology. Results of the full study will provide nationally representative data and ultimately may be used to improve the quality of care, provision of rehabilitation services and inform FDA regulatory approval.

CLASSIFYING AND QUANTIFYING UNILATERAL PROSTHESIS USE IN HOME ENVIRONMENTS TO INFORM DEVICE AND TREATMENT DESIGN

Adam Spiers[1], Linda Resnik[2], Aaron Dollar[1]

1. *Department of Mechanical Engineering, Yale University, USA*
2. *Providence VA Medical Center and Brown University, USA*

BACKGROUND

Although there has been substantial efforts to develop new upper limb prostheses, evaluation of such systems is typically conducted through highly structured tests in clinical / laboratory settings or through survey studies. Though such evaluation techniques provide valuable data, they do not characterize how amputees make use of their prostheses in daily life.

PURPOSE

In our work, we seek to objectively classify and quantify how experienced unilateral upper-limb prosthesis-users utilize their own prosthetic devices and unaffected limbs while completing unstructured and unsupervised manipulation tasks within their own home. Our goal is to identify usage trends in naturalistic everyday activities to inform the design of new prosthetic devices and/or therapeutic interventions.

METHODS

Our analysis is based on 'first-person perspective' video recordings from head-mounted lightweight cameras, which can record for several hours at a time. The cameras are pointed towards the hands and arms of participants, who are given a short list of recommended tasks (e.g. vacuuming, brushing teeth) but mostly complete self-chosen domestic 'housework' activities during data collection periods. To date, we have collected 16 hours of video recordings from 3 participants. Two are congenital transradial amputees (one female) who use body-powered devices and one is a male with shoulder-disarticulation amputation who uses a myoelectric powered elbow, wrist rotator and multi-grasp hand. Classification of the observed manipulation strategies led to the generation of a 'Prosthesis-User Manipulation Taxonomy' which accounts for all observed actions via manipulation 'tags' split into three categories of 'Intact Hand', 'Prosthetic Device' and 'Bi-Lateral' with an additional tag for environmental features use to aid manipulation ('Affordance'). The tags consider both prehensile 'grasping' motions in addition to non-prehensile interactions, such as pushing, leaning, clamping objects against the body or hanging objects from the Terminal Device (TD).

RESULTS

Our preliminary results stem from in-depth 'tagging' of segments of the videos using the taxonomy. We have identified several thousand tag instances at an average of 33 manipulation tags per person, per minute. Though recruitment and video analysis is ongoing, initial observations are that non-prehensile manipulation with the TD occurs significantly more often than prehensile manipulation for participants with transradial amputation. This suggests that device design efforts may benefit from focus on non-prehensile features (such as grip pads on the outside of a TD). Conversely, the participant with shoulder disarticulation completed few non-prehensile motions (which rely on arm mobility) but did utilize his multi-grasp TD to perform more prehensile grasps.

UPPER LIMB MYOELECTRIC PROSTHESES: USER AND THERAPIST PERSPECTIVES ON QUANTIFYING BENEFITS OF PATTERN RECOGNITION CONTROL

[1]Andreas Franzke, [1]Raoul Bongers, [2]Barbara Pobatschnig, [2]Fabian Unglaube, [2]Andreas Kranzl, [1]Alessio Murgia and [1]Corry van der Sluis

[1]University of Groningen, University Medical Center Groningen, Department of Rehabilitation Medicine, Groningen, the Netherlands
[2]Gait and Motion Analysis Laboratory, Orthopaedic Hospital Speising, Vienna, Austria

ABSTRACT

Pattern Recognition (PR) might facilitate intuitive prosthesis control, which in return should facilitate the execution of activities of daily living (ADLs) for amputees. Tests to evaluate PR-based control and its implementation in daily use are currently not available but are required. These tests should ideally reflect realistic prosthesis utilization, and be based on ADLs that users of modern multi degree-of-freedom (DOF) prosthetic hands perceive as relevant or difficult to execute.

The aim of this study was therefore to describe control issues with current multi DOF prostheses and to identify pertinent ADLs from the perspective of patients and therapists.

We conducted semi-structured interviews with 16 adult patients and 7 therapists who were experienced with multi-DOF myoelectric prosthetic hands. Patient inclusion criteria were unilateral amputation at transradial or wrist level and at least six months of experience with the above mentioned prosthetic hands. Therapist inclusion criteria were at least six months of experience with the treatment of such patients. Moreover we included patients and therapists who already had experience with PR-based myoelectric prostheses. The interviews were recorded and transcribed by an independent person. Data was analysed according to a 5-step framework approach based on Familiarization, Creating a thematic framework, Indexing, Charting, and Mapping & Interpretation.

Myoelectric control was often described as too slow, fatiguing and requiring strong mental effort due to non-intuitive mode-switching signals. This also led to the selection of a small number of employed grip/movement modes. Relevant and difficult ADLs differed between individuals, but recurrent domains were mostly preparation of food, eating and dressing. Many patients perceived their multi-DOF prosthesis as a tool which should be able to support the sound hand in bimanual tasks, when these become very difficult or impossible to perform with one hand. Patients mainly choose a multi-DOF myoelectric hand because they expect additional functionality in comparison to conventional myoelectric prostheses, which may not always be experienced ultimately. Persisting problems were low technical robustness and poor manufacturer support (e.g. long waiting times for replacement of parts).

This study revealed several aspects worth considering when testing future (PR-based) myoelectric prostheses. The test should involve bimanual tasks related to preparing food, eating or dressing, where the prosthetic hand works to assist the sound hand. Next to completion time and quality of movement, variables such as ease of use, mental effort, embodiment and grip type variety might indicate whether PR holds benefits for the patient in myoelectric prosthesis control.

MYOELECTRIC PROSTHESIS FOLLOWING TOTAL THUMB AMPUTATION

[1]Vera van Heijningen, [2]Raymon Wijman, [3]Veit Biedermann and [1]Wim Janssen

[1]Rijndam Rehabilitation Institute Location Erasmus Medical Center
[2]Livit
[3]Luttermann

AIM AND OBJECTIVES

In clients with thumb amputation, a passive thumb prosthesis is one possible solution to restore pinch grip and grip to improve hand function. However, the clinical experience of the authors is that this is not routinely a satisfying solution for the client with a total thumb loss.

With the development of Myoelectrically controlled digit prostheses, prosthetic options for partial hand deficits have been widened.

The aim of this study was to investigate the potential benefit of applying a myoelectric controlled thumb in comparison to a passive prosthetic thumb.

This case study presents the process and preliminary results of optimizing the hand function of a client with a total thumb loss.

CASE DESCRIPTION AND METHODS

A 53-year old woman reported impairment in daily activities as her chief complaint.

The thumb of her right hand was amputated at the level of the carpo-metacarpal joint, following a period of severe suffering from Complex Regional Pain Syndrome type II as a result of a cat bite. Her left arm showed severe signs of overuse as a result of the limited capacity of her right hand. This case report describes the process of designing and manufacturing a myoelectric thumb prosthesis. Including the rehabilitation process that followed

FINDINGS AND OUTCOME

This case-report also describes the rehabilitation process which was focused on regaining balanced use and interlimb interaction in daily activities. Experiences, advantages and disadvantages of the three options (no prosthesis, passive prosthesis or myoelectric prosthesis) will be shared and discussed. Patient rated outcome measures show positive results.

IN CONCLUSION

The myoelectric thumb prosthesis restored the hand function beyond the client’s expectations. The client has regained the ability to be fully active in her daily life, in the most practical, comfortable and secure way as possible. Furthermore, her self-esteem and self-image have grown.

The results obtained in this case report do not automatically transfer to other cases. Further research is needed

CLINICAL RELEVANCE

No case-report or any literature on this topic was found by the authors. This case report has identified a potential improvement of hand function for clients with total thumb amputation by using a myoelectric thumb component .

PERCEPTUAL AND CONTROL PROPERTIES OF A HAPTIC UPPER-LIMB PROSTHETIC INTERFACE

Alistair N. Vardy, Manon Boone, and Dick H. Plettenburg

Delft Institute of Prosthetics and Orthotics, Delft University of Technology, The Netherlands
d.h.plettenburg@tudelft.nl; www.dipo.3me.tudelft.nl

ABSTRACT

Current upper-limb prostheses fail to meet user expectation. Common complaints are the large control forces for body-powered prostheses (BPP) and the lack of proprioceptive feedback in myo-electric prostheses (MEP). This study investigates sensitivity and control accuracy and feedback sensitivity of a haptic (master-slave) interface that combines BPP-like control for a MEP utilizing low control forces in the presence of proprioceptive feedback. This first step focuses on static forces. One experiment focused on the just noticeable differences (JND) and the Weber fraction (WF) of the shoulder, and a second on force control. JND and WF were determined by a two-alternative-forced-choice-method at 4 forces levels (2, 4, 6, and 8 N). Force control was evaluated by a visual matching task and blind reproduction task at the same force levels, and force error (FE) and force variability (FV) were obtained. WF results (7% for 2 N, 3% for 4 N, 3% for 6 N and 2% for 8 N) indicated a level of sensitivity comparable to human weight perception. FE and FV values were small enough as not to affect usability when grasping objects. We conclude that forces of 2-10 N are sufficient to operate an externally powered prosthesis while maintaining a sufficient level of proprioceptive feedback.

INTRODUCTION

Despite advancement in prosthesis design, many users remain dissatisfied [1], where the main complaints concern discomfort of wearing the prosthesis, the difficulty to don and doff, and the difficulty of controlling the prosthesis which demands a high mental and physical load [1]. When considering the two main types of upper-limb prostheses, body powered prostheses (BPP) and Myo-electric prostheses (MEP), BPP suffer from high control forces resulting in discomfort and compensatory movements, and MEP are accompanied by a high mental load due to the lack of proper feedback [2]. A good prosthesis should provide the user with enough information to perform daily tasks that require fine motor adjustments. This can only be achieved if there is sufficient (proprioceptive) feedback [2, 3].

This paper takes a first step in investigating a new method for controlling an upper limb prosthesis combining the strengths of both BPP (proprioceptive feedback) and MEP (low control forces). More specifically, we aim to develop a haptic interface that provides proprioceptive feedback and allows the user to set a comfortable control force range and feedback for optimal gripping experience. The system consists of two parts, a master and a slave system. The master system is worn on and controlled by the shoulder. The slave side is an externally powered prosthesis.

Improvements in cosmesis were obtained by using skin anchors instead of a shoulder harness [4]. The use of skin anchors also removes the discomfort of the harness digging into the skin. Compared to BPP, control forces are markedly lower. Pilot studies revealed that static forces up to 10 N were most comfortable, and forces below 2 N were deemed too low for useful control.

To improve feedback, the master device provides proprioceptive feedback to the user. As the control system is implemented electronically, the control and feedback level can adjusted to suit the user. By using two skin anchors [4], the device can easily fit underneath clothing. Note that, the shoulder-worn master system can be placed on the affected and healthy side of the body. See also Figure 1. Control is improved by using control forces up to 10 N. Together, the proposed solution provides the necessary level of comfort, control and cosmesis to the user [3].

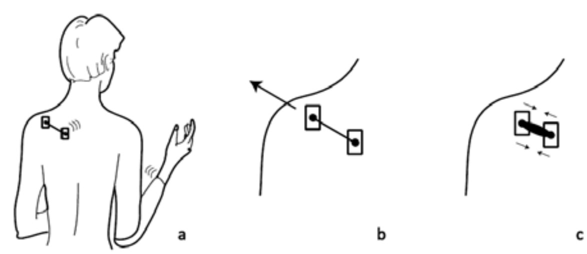

Figure 1. The intended solution for the new prosthesis interface. The control forces are exerted by the same movement as a body-powered prosthesis (a). Force is measured by a sensor between the skin anchors (b) and proprioceptive feedback is provided by pulling the two skin anchors closer together (c).

Specifically, this study investigates the sensitivity of a dual skin anchor solution for static forces by determining the Weber Fraction (WF) and the Just Noticeable Difference (JND) for forces and comparing the results to known perceptual results for weight sensitivity. Weber's Law predicts a subject's difference threshold of any intensity value, and the WF equivalent is defined as $\Delta\Phi/\Phi=c$, where

$\Delta\Phi$ is the difference threshold or JND, Φ the initial stimulus intensity and c is a constant; the WF [5, 6]. The same law also states that the JND is proportional to the original stimulus. The WF is a ratio comparing the stimulus intensity with the change in the stimulus magnitude [5]. The WF shows the minimum decrease or increase an object has to make in order to be noticeable. A WF of 0.05 or 5% means that a subject can reliably detect a change of 5% in stimulus intensity. In addition we will investigate the relationship between applied and perceived force by performing a force reproduction experiment.

These two experiments will be performed and the following question will be answered in experiment 1: 'what is the Weber fraction equivalent for the threshold of force feedback for the skin anchors used for prosthetic control?'. Experiment 2 will answer the question 'what is the accuracy of force control, in terms of force perception and force reproduction, for shoulder muscles used in this type of prosthetic control?'

METHODS

Two experiments were performed. The first experiment focused on the WF for static force feedback and the second focused on the accuracy of control where subjects followed a force profile. Both experiments used the right shoulder to produce force by elevations and protraction. Forces were sufficiently low as not to induce fatigue. Both experiments were performed in a single session separated by a break and lasted 1.5 hours. The same group of subjects participated in both experiments. Both experiments used the same setup.

Participants

Ten healthy, right-handed males aged 18 to 37 performed in both experiments. Subjects provided informed consent and both experiments were approved by the local ethics committee.

Setup

Forces were recorded by a load cell (Futek LSB200, 50 lbs) amplified (Scaime, CPJ), discretized (NI, MyDAQ) and stored using custom Matlab (R2014b) software. A load cell interrupted a cable connecting two skin anchors positioned on the right shoulder and to the right of vertebrae T7, see Figure 2.

Methods experiment 1 – The Weber fraction equivalent

The 2AFC discrimination task is a psychophysical method and is commonly used to measure performance as a proportion of correct responses when comparing two stimuli. We determine if the subject judged test force to be larger than a reference force [5]. Perception is determined during active force production by the subject to simulate normal prosthetic use.

Figure 2. The experimental setup for both experiments

Each session started with several familiarization trials. Four reference forces were used: 2, 4, 6, and 8 N. Trials for the different reference forces were blocked and the order of the reference forces was counterbalanced.

A trial consisted of the subjects exerting force and matching a reference force level and a test force level (the order was randomized). Each force was produced for 5 seconds with a 3 second delay between the two force levels. In between forces, the subject was asked to relax and prepare by cues on the screen. Target height was kept constant to eliminate any visual cues relating to the magnitude of the target force. The target ensured that force level was not simply discernible from the magnitude of the movement. The test force levels comprised 10 stimulus intensities deviating from the reference force at ±3.5%, ±7%, ±10.5, ±14%, and ±17.5%. Each of the 10 test force levels was presented 4 times for each of the 4 reference forces yielding a total of 160 trials. The 160 trials took about 1 hour to complete.

Data analysis – experiment 1

During the experiment the number of trials where the test force was identified as larger than the reference force were counted and divided by the number of repetitions (4). Data for all subjects were pooled and a logistic psychophysical curve was fitted for each of the references forces (Psignifit 3.0, bootstrap method http://psignifit.sourceforge.net/). The JND was determined by determining the differences in force (ΔF) corresponding to 25% and 75% success probability using the following formula [7]:

$$JND = (\Delta F(75\%) - \Delta F(25\%))/2 \quad (1)$$

The calculated JNDs were used to determine the WF by dividing the JNDs by the different reference forces. A low

WF indicates a high level of precision. The WF was determined using the following formula:

$$WF= JND/F\_ref \times 100\% \qquad (2)$$

Methods experiment 2 – accuracy of force control

The second experiment consisted of 4 blocks of 10 trials each corresponding to the four reference forces (2, 4, 6, and 8 N). The order of the 4 blocks of the 4 reference forces was counterbalanced. Each trial consisted of producing a constant force level with visual feedback, where a line on a screen indicated the target force. This force was exerted for 5 seconds. After completion, the subject was asked to reproduce the force level without visual feedback, also for 5 seconds. The two stimuli were separated by 3 seconds and relax/starting cues were displayed on the screen. Data was recorded with a sampling rate of 1 kHz. The 40 trials of experiment 2 took about 20 minutes to complete.

Data analysis – experiment 2

From each trial, the first 2.5 seconds and the last 0.5 seconds were discarded to remove transition effects. The remaining samples were analysed to obtain two measures: 1) the force error (FE), which was defined as the absolute difference between mean of the data and the target, 2) the force variability (FV), which was defined as the standard deviation of the produced force.

RESULTS

The pooled results revealed JND values of 0.14, 0.11, 0.16 and 0.17, and WF of 7%, 3%, 3%, and 2% for 2, 4, 6, and 8 N respectively. Figure 3 shows the psychometric functions indicating the probability of a correct response as a function of the difference between the forces of the reference and the test stimuli. Higher differences between force levels were detected correctly more frequently than lower differences.

The FE (Figure 4) and FV (Figure 5) results for the force reproduction task are presented below. Error bars represent the standard error of the mean. FE is markedly lower in the presence of visual feedback compared to the blind reproductions trials. Both conditions, however, illustrate the know property where the error scales with force level, as also seen in the FV. Surprisingly, the FV was comparable for both conditions. The FV shows the biggest variation for 8 N, with a mean variability of about 0.2 N.

DISCUSSION

The first experiment investigated sensitivity for static force production using shoulder movements. In summary, we found good force discrimination using shoulder movements with WFs: 7% for 2 N, 3% for 4 N, 3% for 6 N and 2% for 8 N, which are in the range of weight perception [5]. However, for 2 N, the sensitivity was markedly lower.

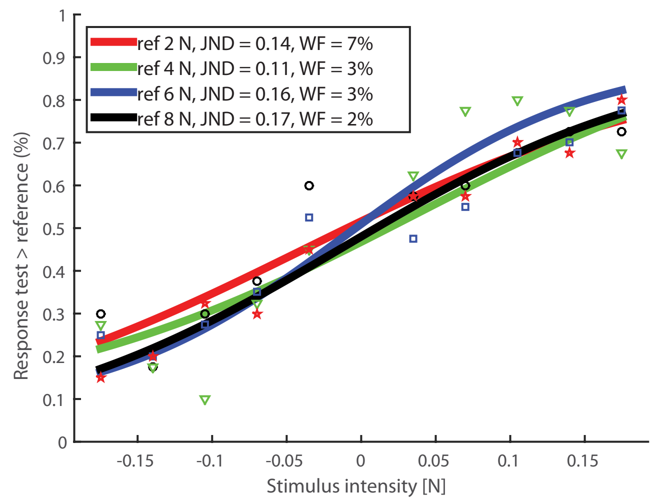

Figure 3. The psychometric functions for the four reference forces.

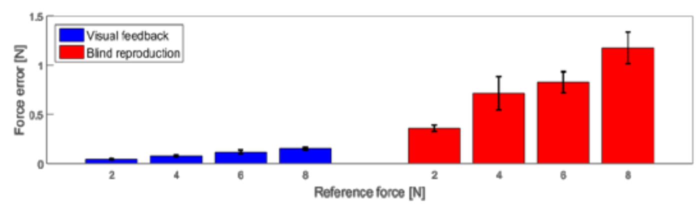

Figure 4. The FE averaged over subject and repetitions for the visual matching and blind reproduction tasks.

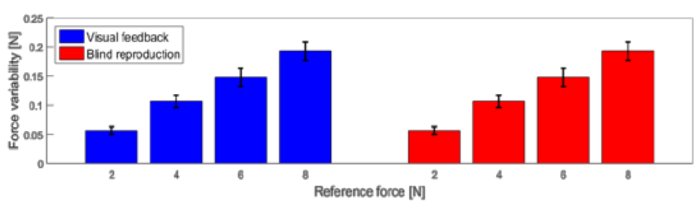

Figure 5. The FV averaged over subject and repetitions for the visual matching and blind reproduction tasks.

The resulting WFs ranged from 2% for 8 N to 7% for 2 N. The WF of 2% for 8 N shows that the subjects have a high level of sensitivity for differences in force levels. Previous studies on WF for force detection in healthy shoulders found fairly similar outcomes. Feyzabadi et al. [6] used also a 2AFC paradigm where the subjects had to perform a rotating shoulder movement in the upper arm for 3 different force intensities of 0.5, 1 and 1.5 N, and found a WF for the shoulder of 8% [6]. Hurmuzlu et al. [8] used shoulder-elbow motions to determine the JND for masses of 1, 2, 3 and 4 lb, and found WFs of 5%, 1.25%, 17% and 6%

respectively. Our results suggest that control using two skin anchors has a more than adequate level of sensitivity.

The focus of the overall project is to decrease the forces required to operate a prosthesis while still having a good level of proprioceptive feedback. The WF for different forces levels indicate that this has been achieved. Prosthetic users tend to prefer body-powered prostheses over externally powered prostheses because of the possibility to receive proprioceptive feedback, which is not available with externally-powered prostheses [3]. The results from Experiment 1 indicate that skin anchors may be intuitively used to control an externally powered prosthesis with shoulder movements.

The second experiment was designed to separate the influences of force perception and force reproduction on the judgment of the different force levels. Force reproduction gives insight in force production and force perception. Both force production and force perception are of interest for the control mechanism of the newly developed prosthesis. When it is able for a future user to produce an intended, or previously perceived force, the control over the artificial hand will increase. In summary we found that during the reproduction test the FE was increased compared to visual matching, this is in agreement with previously found data [9, 10]. This increase in FE is largest for the reference force 8 N, with a FE of approximately 1.2 N.

The results of the second experiment show that the force level affects FE and FV, where both measures increase when force levels increase; a known property of muscular control [9]. The FV seems independent of feedback as both the visual matching test and the blind reproduction test provide us with similar values. The FV was measured to give information about the force control and seems to be comparable, for inexperienced subjects using, in both visual and blind tests. Force errors did not exceed about 1 N and variability was less than 0.2 N. These results indicate that a user would not drop or crush an object.

One of the main advantages of this system is the presence of mode-specific proprioception, i.e., proprioception is provided for the muscles that provide force. This type of control is natural to the human body and is the fastest and most intuitive form of feedback and affords a reduced mental workload for the user [2].

The control source was chosen to be the shoulder as degree of freedom that could be manipulated without changing the position of the prosthesis. For example, elbow-powered prostheses require elbow flexion and extension to open and close the prosthesis resulting in displacement of the object that is gripped. With shoulder control, however, an object can be held at any position within the range of motions while still being able to change grip force.

The main motivation for using skin anchors was to provide an unobtrusive solution which could be hidden beneath clothing. We believe that the final design of the complete system, (anchors, actuator, power source) will remain small enough to achieve this goal.

CONCLUSION

The results presented in this paper indicate that a haptic interface using two skin anchors may provide a solution that allows low control force in the presence of meaningful force feedback. This study was a first step and efforts are underway investigating dynamic grasping. Pilot results indicate that this system performs as well as BPP for the Box and Blocks test [11].

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DEVELOPMENT OF THE HANDI HAND: AN INEXPENSIVE, MULTI-ARTICULATING, SENSORIZED HAND FOR MACHINE LEARNING RESEARCH IN MYOELECTRIC CONTROL

Dylan J. A. Brenneis[1,4], Michael R. Dawson[2,4], Patrick M. Pilarski[3,4]

[1]Department of Mechanical Engineering, University of Alberta; [2]Glenrose Rehabilitation Hospital, Alberta Health Services; [3]Department of Computing Science, University of Alberta; [4]Division of Physical Medicine and Rehabilitation, University of Alberta, Edmonton, AB, Canada

ABSTRACT

Machine learning (ML) has been applied in both research and clinical settings to make myoelectric prostheses more functional and more intuitive to use. ML techniques for myoelectric control require information about the environment a control system occupies in order to make useful control decisions or predictions about a user's desired control outcomes. Despite demonstrated increases in myoelectric control performance with the inclusion of additional information about users and their environments, the sensors in commercial prostheses are limited, and typically do not provide diverse channels of contextual information to their respective control systems. Additional sensor information is crucial to demonstrating and evaluating the full potential of next-generation ML control systems. With this in mind, a novel, cost-effective research prosthesis was designed to provide real-time sensory information for ML-based myoelectric control. This device is able to report fingertip forces on independently controlled fingers, angular position for individual finger joints, and visual information about the hand's environment via a USB webcam integrated in the palm. Using 3D printing, the device was prototyped at a cost of less than $800 CAD. This work therefore contributes a new platform by which groups can conduct ML research on prostheses, and allows researchers to develop new ML approaches with ample access to contextual information about prosthesis movement, prosthesis-environment interactions, and local changes to the environment surrounding the prosthesis. By providing an inexpensive, highly sensorized prosthetic hand, this work helps mitigate the cost of purchasing and retrofitting commercial prostheses with new sensors; it is therefore also expected to support related research into methods for sensory feedback from prostheses to users.

INTRODUCTION

Advanced bionic limbs such as the Modular Prosthetic Limb (Johns Hopkins University, Laurel MD, US), Bebionic (RSL Steeper, Leeds, UK), or i-Limb (Touch Bionics, Mansfield MA, US) demonstrate increasing similarity to biological limbs in terms of their functionality and their kinematic capabilities. However, as with many systems that have more degrees of freedom than available control signals, the in-practice dexterity of myoelectric prostheses is limited by the effort required to control them—non-intuitive control and lack of accessible functionality are two of the main reasons for the low acceptance rate of upper-limb myoelectric prostheses [1–3]. Machine learning in the form of pattern recognition, regression learning, and reinforcement learning have all been demonstrated as ways to potentially reduce the control burden on users while increasing the functionality of prosthetic devices [2].

Previous work has shown that increasing the sensor space provided to a pattern recognition system—for example, adding accelerometers to provide a sense of residual limb position—have a noticeable effect on the ability of the control system to make correct classifications in different situations [4]. More generally, it is natural to expect that machine learning predictions, and therefore control decisions, may be improved by increasing the control system's awareness of the environment it is acting in—that is, by giving the learning system increased sensory feedback. Sensory feedback has also been shown to improve a user's control over their prosthesis [5]. Commercially available prostheses however, either lack the ability to gather sensory information, or lack a purposely-designed means of transferring detailed information to the user or to a learning agent [6]. Further, the cost of these devices can be prohibitive in some research settings.

By improving a prosthetic control system's window into its own operation, the world around it, and the intentions of its human user, significant gains can be expected in terms of the capacity of that system to meets its user's needs in diverse and changing circumstances [2, 7]. The present work therefore contributes a novel research prosthesis—the Humanoid Anthropometric Naturally Dexterous Intelligent (HANDi) Hand—that has been designed to allow a rich stream of information to be delivered to a machine learning agent or human user.

Table 1: Design specifications for HANDi Hand.

Item	Design Specification	Achieved Spec.
Size	Full scale, anatomical proportions	Criteria met
Mass	< 500 g	256 g
Max Payload	500 g	500 g
Degrees of Freedom	Flexion/extension of each finger plus abduction/adduction of thumb	Criteria met
Degrees of Actuation	Each finger independent; thumb adduction separate from flexion	Criteria met
Sensing	Position, fingertip force, visual data	Criteria met
Interface	Compatible with Bento Arm	Criteria met
Modularity	Exist as standalone system	Criteria met
Prototype Cost	< $2500	$800
Finger Speed	Full close in < 1 s	0.43 s
Grip Force	> 4 N cylinder grip	4.2 N

DESIGN SPECIFICATIONS

The design specifications for the HANDi Hand are outlined in Table 1. The device was designed to operate either as a standalone system or in conjunction with the Bento Arm—a 3D printed upper-limb prosthesis previously developed at the University of Alberta for myoelectric training and research [8]. As such, the size was specified to be 1:1 scale with anatomical proportions similar to the Bento Arm. In order to not detract significantly from the Bento Arm's total payload capacity, the mass of the device was restricted to less than 500 g. Most tasks performed with the device for machine learning trials and many tasks of daily living do not require large payloads; therefore the hand was specified to support a maximum payload of 500g in a cylinder grip (equivalent to holding a 500 mL water-bottle). This equates to approximately a 4 N cylinder grip.

To ensure natural dexterity of the device, all natural degrees of freedom of a human hand were included in the design, with the exception of lateral finger movement. These degrees of freedom were excluded due to the increased level of complexity they would introduce. Many commercially available hands underactuate the fingers, causing fingers to flex and extend in concert with one another to simplify control. Since the goal of the research is to make control intuitive without loss of dexterity, it was decided that each finger should be independently actuated, and thumb rotation should be independent of thumb flexion.

Key sensory abilities to include were force and position sensing for each of the fingers, as these would provide similar information to what a biological hand would naturally supply. To allow exploration of non-physiological sensory capabilities, the capacity for visual feedback was also included via an in-palm camera.

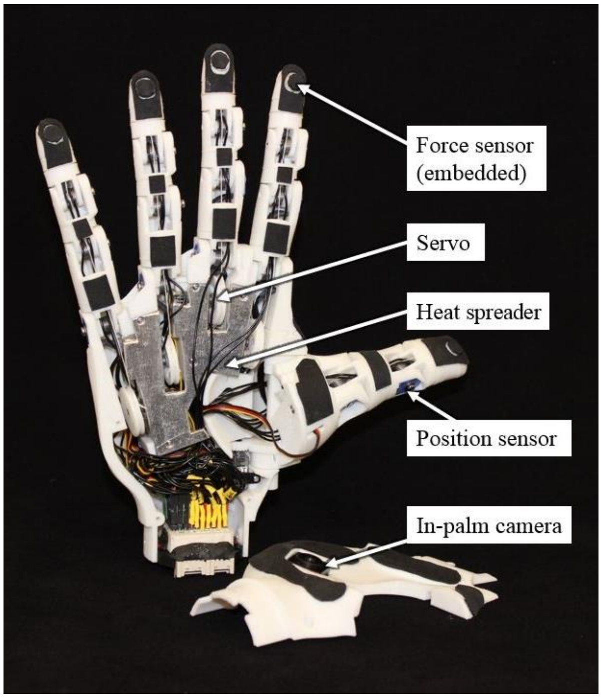

Figure 1: Assembled HANDi Hand prototype with six servo motors installed and palm cover removed.

In order to make the device accessible to other research groups and facilitate rapid design iterations, the cost of the device was specified to be less than $2500.

MECHANICAL DESIGN

All parts of the device were modelled in 3D design software and printed on a Replicator 2 desktop 3D printer (MakerBot Industries, LLC, Brooklyn NY, US). As a starting point, the finger was modelled after a previously designed open-source finger from the InMoov project [9]. Modifications were made to this finger to allow for the introduction of position sensors, force sensors, and an altered extension scheme. The thumb was then modelled using the same hinge mechanism with modified proportions. In order to maintain anthropometric dimensions, the size of the fingers, thumb, and carpus were modelled to match a 50^{th} percentile male. Each of the fingers are identical in length, modelled after the proportions of the ring-finger, in order to simplify design iterations. Anthropometric considerations of individual finger length are compensated for by unique positioning of each finger on the carpus. The assembled prototype, with palm cover plate removed to show interior workings, is shown in Figure 1.

Flexion of each finger is actuated by the rotation of a Hitec HS 35-HD servo motor. These motors were chosen because their small size allows all components to fit within the palm of the hand, as per the modularity requirement.

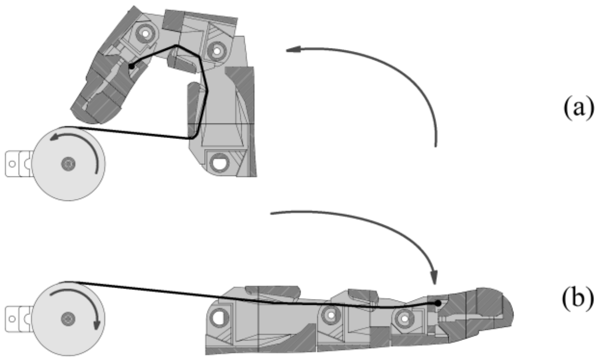

Figure 2: Cutaway view of finger actuation mechanism showing (a) flexion by winding up on the servo spool and (b) extension by releasing the spool allowing torsion springs at each joint to extend the finger.

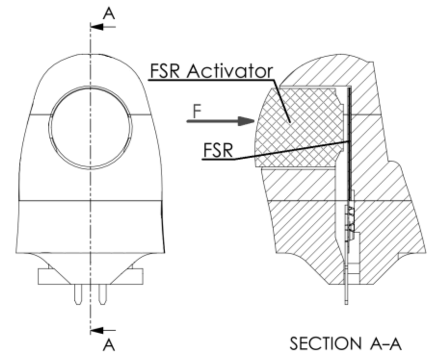

Figure 3: Section view showing FSR setup. Force F displaces the FSR activator column, thereby applying force to the embedded FSR.

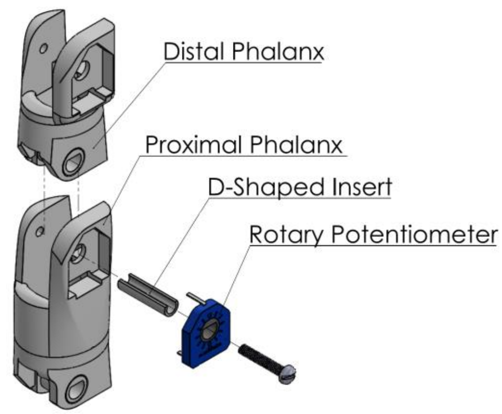

Figure 4: Exploded view of one finger joint showing potentiometer setup. The D-shaped insert rotates with the distal phalanx, turning the hub of the potentiometer. The potentiometer is fixed relative to the proximal phalanx.

As the servo horn turns, a nylon thread attached at the tip of the finger spools around the servo horn causing flexion as seen in Figure 2. As the servo rotates in the opposite direction, the thread unspools and a torsion spring at each joint causes extension. This mechanism allows the finger to wrap around objects using the distal joints.

During long periods of continuous use, the temperature of the servomotors can reach temperatures high enough (~85˚C) to cause the 3D printed plastic to soften. To combat this, a plate was machined from 2 mm thick aluminium and acts as both a heat spreader and cover plate keeping the servos in place. The wrist of the hand is designed to mount directly to the Bento Arm, and can also be mounted to a separate stand for independent use.

ELECTRICAL DESIGN

Control of the servos and aggregation of sensor signals is accomplished on an Arduino Mega (Arduino LLC, Italy), chosen due to its high number of analog pins. Servo position is encoded using velocity control over the Arduino's PWM outputs. To negate voltage drop on sensor readings, the servos are powered separately from the rest of the system. The servos can be controlled by any analog signal: for example, joystick or myoelectric signals. When used with the Bento Arm, the Arduino takes control signals from and sends sensor data to a BeagleBone Black Rev2, which operates on the Robot Operating System (ROS).

Force sensing at each fingertip is accomplished by use of a force sensitive resistor (FSR). Change in pressure normal to the FSR causes a proportional change in its resistivity and therefore voltage drop, which is measured through a voltage divider circuit. The FSRs are embedded within the fingertip to ensure consistent orientation regardless of change in the environment, and are actuated by pressing on a column that runs through the fingertip normal to the FSR (Figure 3).

The angle between each joint is measured by a MuRata SV series rotary potentiometer (MuRata Manufacturing Co. Ltd., Kyoto, JP). A D-shaped plastic insert rotates with the more distal phalanx while the body of the potentiometer is fixed to the more proximal phalanx as pictured in Figure 4. Due to limitations in the number of analog signal pins available on the control board, the most distal joints in each finger as well as the intermediate joints in the little and ring fingers are not equipped with potentiometers at this time; allowances have been made for their addition in the future.

An in-palm USB webcam was included by dissecting a Logitech QuickCam Pro (Logitech, Newark CA, US), and appropriating the functional components for use in the hand. The camera's circuit board and lens is attached to a custom mount on the palm cover, and the original USB cable passes directly out of the hand and into the computer, bypassing the control board.

PROTOTYPE CHARACTERISTICS

Force output from the fingertips was measured using an external load cell (LCM703-5, OMEGA, Laval QB, CA, calibrated to an accuracy of 0.02 N, resolution 0.003 N) placed at the fingertip approximately 77 mm from the metacarpophalangeal (MCP) joint. The average force output was 0.81 N, s = 0.05 N. The grip force was measured using the same load cells mounted inside an 80 mm diameter cylinder. The hand provides a radial cylinder grip force of 4.21 N, s = 0.33 N. In practice, the hand was able to support a 500 g water-bottle without slipping.

Embedding the FSR within the fingertip limits the perception of force to a binary indication of the presence of force. The sensor reading increases with greater forces, but the differences were not found to be significant enough to establish a relationship. There is a measurable difference between the FSR voltage at no force applied and at forces greater than 0.20 N, indicating that the sensors provide a reliable indicator of whether or not force is applied.

Repeatability of the finger movement was measured using the built-in potentiometers. The potentiometers (accuracy 1.2°, resolution 0.35°) were calibrated within the finger using a goniometer. In both open and closed positions, the angle of joints with potentiometers were analysed, and found to be repeatable with a standard deviation of less than 3.4° across all measured finger joints.

Material cost of the prototype including all hardware, sensors, and servos was less than $800 CAD, not including shipping charges. Fabrication and assembly costs are also not included in this appraisal.

FUTURE WORK & CONCLUSIONS

In order to make the HANDi Hand a fully modular system, control and signal acquisition should be integrated into the cavity of the palm rather than existing as it does now on an external Arduino. This will require custom PCB design, which will increase the cost of manufacturing and may create a barrier to other groups wishing to build their own version. Integration of control into the hand will however reduce the amount of cables required, making the hand more accessible for wearable use. Further improvements to the current prototype will involve more sensing capabilities: load sensing, temperature sensing, additional cameras, and more comprehensive position sensing are near-term considerations. Also being considered is the inclusion of an LCD display screen for visual feedback to the user.

In conclusion, the HANDi Hand is a functional prototype with force, position, and visual sensory capabilities that can be constructed for less than $800. The sensory information it provides will enable machine learning control systems to more accurately represent the hand's environment and interactions. Increased sensory information to a machine learning myoelectric control system is expected to enable contextually appropriate control decisions on the part of the controller, and more appropriate natural myoelectric control for end users.

ACKNOWLEDGEMENTS

The authors would like to acknowledge Jaden Travnik for his work on the software integration of the HANDi Hand with Bento Arm, and Jason Carey for conversations and helpful suggestions relating to this manuscript. This work was made possible thanks to support from the Alberta Machine Intelligence Institute (Amii), Alberta Innovates – Technology Futures, the Government of Alberta, the Natural Sciences and Engineering Research Council (NSERC), and the Canada Research Chairs program. DJB was supported by an NSERC USRA and a University of Alberta Undergraduate Research Stipend.

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DESCRIPTIVE OUTCOME METRICS OF SENSORIZED UPPER LIMB PERFORMANCE USING OPTIMAL FORAGING THEORY

[1]Dylan Beckler, [1,2]Zachary Thumser and [1,3]Paul Marasco

[1]Laboratory for Bionic Integration, Department of Biomedical Engineering, Lerner Research Institute, Cleveland Clinic
[2]Research Service, Louis Stokes Cleveland Department of Veterans Affairs Medical Center
[3]Advanced Platform Technology Center of Excellence, Louis Stokes Cleveland Department of Veterans Affairs Medical Center

ABSTRACT

Modern advancements of upper-limb prosthetic technologies have not been accompanied by advancements in appropriate metrics for assessing the functionality of these technologies. In particular, many of the currently accepted functional metrics of performance for upper-limb prostheses put little or no emphasis on the role of sensory feedback modalities in the prosthesis control loop. We developed a functional metric of Prosthesis Efficiency and Profitability (PEP) which incorporates tactile and proprioceptive elements into a simple motor task. PEP uses Optimal Foraging Theory (OFT), which describes decision making in biological systems based on time versus prey-value tradeoffs, as a platform to evaluate the compensatory interactions between motor command and sensory feedback in a prosthesis control loop. PEP participants are instructed to discriminate between objects of different stiffnesses in a timed search and acquisition task. The primary outcome measures: efficiency and profitability, weigh the accuracy of stiffness discrimination against speed. Additionally, Bayesian statistics are used to determine the frequency of false positive and false negative errors made by the participant during the test to further describe the tradeoff between motor command and sensory feedback. We have found that the PEP test is sensitive to the effects of touch and movement feedback and highlights strategy switches and changes in performance for different devices and feedback settings. We observed subjects switching from relying on motor command to relying on sensory feedback when the feedback was turned on. When feedback was absent, participants on average had lower accuracy but compensated by engaging with objects more quickly. Whereas when feedback was present, participants tended to spend more time engaged with each object but got more objects correct. Additionally, an ability to discriminate stiffnesses beyond chance was demonstrated in prostheses users equipped with touch and movement feedback when the sensory feedback was switched on. The PEP test may provide a general framework for evaluating different sensory modalities; the objects of different stiffnesses may be replaced with objects possessing the property of interest, changing the focus of the test without altering the analysis or interpretation of outcome metrics.

RELIABILITY IN EVALUATOR-BASED TESTS: A MODELLING APPROACH FOR INTERPRETING INDICES OF RELIABILITY AND DETERMINING AGREEMENT THRESHOLDS

[1]Dylan Beckler, [1,2]Zachary Thumser and [1,3]Paul Marasco

[1]Laboratory for Bionic Integration, Department of Biomedical Engineering, Lerner Research Institute, Cleveland Clinic
[2]Research Service, Louis Stokes Cleveland Department of Veterans Affairs Medical Center
[3]Advanced Platform Technology Center of Excellence, Louis Stokes Cleveland Department of Veterans Affairs Medical Center

ABSTRACT

Indices of inter-evaluator reliability are used in many fields such as computational linguistics, psychology, and medical science; however, the interpretation of resulting values and determination of appropriate thresholds lack context and are often guided only by arbitrary "rules of thumb" or simply not addressed at all. Our goal for this work was to develop a method for determining meaningful interpretation of values, thresholds, and reliability based on a systematic alteration of the mean and variance within a normally distributed error signal, providing insight into the interplay between bias and error of a hypothetical rater population. As a basic metric for inter-rater reliability we selected Krippendorff's alpha. This is a versatile statistical tool for quantifying the agreement between multiple evaluators on sets of observations or measurements and it is highly flexible in handling multiple raters, missing data, and different scales of measure. We presented a video analysis task to three expert human evaluators and averaged their results together to create an initial dataset of 300 time measurements. We developed a mathematical model that then introduced a unique combination of systematic error and random error onto the original evaluator dataset to generate 4800 new hypothetical raters (each with 300 time measurements).We calculated the percent error and Krippendorff's alpha between the original dataset and each new modified dataset to determine the value envelope of inter-rater agreement. We then used this information to make an informed judgement of an acceptable threshold for Krippendorff's alpha within the context of our specific test. As a marker of utility we calculated the percent error and Krippendorff's alpha between the initial dataset and a new cohort of trained human evaluators, using our contextually derived Krippendorff's alpha threshold as a gauge of evaluator quality. We found that this approach established threshold values of reliability, within the context of our evaluation criteria, that were far less permissive than the typically accepted "rule of thumb" cutoff for Krippendorff's alpha. This procedure provides a less arbitrary method for determining a reliability threshold and can be tailored to work within the context of any reliability index.

CONTROL OF ISOMETRIC GRIP FORCE, VISUAL INFORMATION PROCESSES, AND FITTS' LAW

[1,2]Zachary Thumser, [3]Andrew Slifkin, [1]Dylan Beckler and [1,4]Paul Marasco

[1]Laboratory for Bionic Integration, Department of Biomedical Engineering, Lerner Research Institute, Cleveland Clinic
[2]Research Service, Louis Stokes Cleveland Department of Veterans Affairs Medical Center
[3]Department of Psychology, Cleveland State University
[4]Advanced Platform Technology Center of Excellence, Louis Stokes Cleveland Department of Veterans Affairs Medical Center

ABSTRACT

Fitts' law models the relationship between the amplitude, precision, and speed of rapid movements. It has been widely used to quantify performance in pointing tasks, particularly for Human-computer interaction, but the same model can be applied to analogous tasks. If Fitts' law also applies to grip forces, model parameters would provide a meaningful approach to quantify grasp performance for rehabilitative medicine and prosthetics. We examined the applicability of Fitts' law to a grip force production task, with and without visual feedback (via a force meter), and with target forces presented both explicitly (arrows on the force meter) and implicitly (images of objects). When visual force feedback is available, speed and accuracy of grip force follows Fitts' law (average r2 = 0.82). Without vision (operating exclusively on tactile feedback), accuracy of grip force remains high, but force precision is lower, resulting in overall performance that is relatively insensitive to the target presented. Replacing explicit-but-abstract force targets with images of familiar objects that serve as implicit, well-understood targets enabled participants to generate consistent grip forces more reliably. Population means show that the underlying behavior is well-described by Fitts' law with either vision (r2 = 0.96) or implicit targets (r2 = 0.89), but not for explicit targets without vision (r2 = 0.54). Implicit targets allow for a straightforward and realistic see-object-squeeze-object test that uses Fitts' law to quantify the relative speed-precision relationship of any given grasper.

FUNCTIONAL KINESTHETIC PERCEPTION OF COMPLEX BIONIC HAND MOVEMENTS

[1,2]Paul Marasco, [3,4]Jacqueline Hebert, [5]Jon Sensinger, [1]Courtney Shell, [6]Johnathon Schofield, [7]Zachary Thumser, [8]Raviraj Nataraj, [1]Dylan Beckler, [4]Michael Dawson, [5]Dan Blustein, [5]Satinder Gill, [1]Rafael Granja-Vazquez, [8]Jason Carey and [1,9]Beth Orzell

[1]Laboratory for Bionic Integration, Department of Biomedical Engineering, Lerner Research Institute, Cleveland Clinic
[2]Advanced Platform Technology Center of Excellence, Louis Stokes Cleveland Department of Veterans Affairs Medical Center
[3]Division of Physical Medicine and Rehabilitation, University of Alberta
[4]Glenrose Rehabilitation Hospital, Alberta Health Services
[5]Institute of Biomedical Engineering, University of New Brunswick
[6]Department of Mechanical Engineering, University of Alberta, Mechanical Engineering Building
[7]Louis-Stokes Cleveland Department of Veterans Affairs Medical Center
[8]Department of Biomedical Engineering, Stevens Institute of Technology
[9]Prosthetics and Sensory Aids Service, Department of Physical Medicine and Rehabilitation, Louis Stokes Cleveland VA Medical Center

ABSTRACT

Clinical translation of advanced upper limb prostheses is limited because they do not provide meaningful movement sensation and require constant visual monitoring to complete even the simplest of tasks. Kinesthesia, the sense of body movement, allows us to feel the activity of our extremities without looking at them. This study moves prosthetic feedback into a new perceptual/cognitive framework by harnessing the kinesthetic illusion to provide relevant input to human amputees about complex prosthetic hand movements. In this study, illusion-inducing vibration of surgically reinnervated residual limb muscles in amputees with targeted reinnervation provided physiologically relevant kinesthetic sensation that allowed them to accurately sense and simultaneously control both virtual and mechatronic robotic hand movements in real-time without vision. On a proprioceptive motor task without vision the amputee study participants performed indistinguishably from able-bodied. Psychophysical evaluation of an active motor grip task shows that the kinesthetic feedback alone provides better system resolution than vision alone and when provided with both vision and kinesthesia together they perform optimally. The kinesthetic feedback provided a sense of authorship (agency) over movements and was implemented in clinically realistic 2-site antagonistic myoelectric prosthesis control to provide real-time sensation of hand open and hand close. The feedback system was implementable in physical devices in the context of clinical fitting constraints and the movement percepts can be driven to operate on speed scales relevant to commercially available prosthetic hands. These results open a new path to perceptually-integrated bi-directional bionic prostheses.

USER TRAINING FOR PATTERN-RECOGNITION BASED MYOELECTRIC PROSTHESES USING A SERIOUS GAME

Morten Bak Kristoffersen, Andreas Franzke, Alessio Murgia, Corry van der Sluis and Raoul Bongers

University of Groningen, University Medical Center Groningen, Department of Rehabilitation Medicine, Groningen, the Netherlands

ABSTRACT

Individuals with upper-limb deficiency who are fitted with a prosthesis are normally trained in the use of such device. This is even true for individuals who are fitted with a myoelectric prosthesis that uses control algorithms based on pattern-recognition, despite the intent of pattern-recognition control of exploiting “intuitive” phantom movements. Conventionally, training individuals for pattern-recognition control usually involves an expert who guides the user to produce electromyogram (EMG) signals that optimize pattern recognition. In the training the individual is stimulated to adapt their EMG signals as to make them more distinct in terms of the resulting patterns. To achieve this, for instance, small movements can be added to the basic pattern, such as flexing the little finger during open hand. Although training improves online accuracy it still involves considerable trial and error. Moreover, expert guidance is currently done based on visual perusal of EMG patterns or features thereof and not based on specific metrics characterizing those EMG signal patterns. Rather than using intuitive phantom movements for control, we instead propose to use those phantom movements which are most distinct in terms of EMG. To find the set of phantom movements that provides the most distinct EMG activation patterns, we propose to use a serious game. Using a game, we can train individuals to make EMG patterns distinct while performing them in a robust manner. This game is controlled using the EMG captured from 8 electrodes positioned around the forearm. Inspired by the work of Radhakrishnan et. al and Pistohl et. al, the EMG from each electrode is mapped to a direction of the game avatar in the 2D environment. We hypothesize that this training will make individuals utilize their EMG activation space to a greater extent and become better at generating only EMG activity at specific electrode sites so that patterns are more distinct.

We are currently conducting an experiment in which 4 experimental groups receive different kinds of training. Group 1 receives conventional training without coaching. Group 2 receives conventional training with feedback. Group 3 receives training with the proposed serious game and group 4 receives training without any feedback (control). The learning effects between groups are analysed using the metrics proposed by Bunderson et al. and the motion test.

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INFLUENCE OF A TRANSRADIAL AMPUTATION ON NEUROMUSCULAR CONTROL OF FOREARM MUSCLES

[1]Morten Bak Kristoffersen, [1]Andreas Franzke, [2]Alessio Murgia, [2]Raoul Bongers and [1]Corry van der Sluis

[1]University of Groningen, University Medical Center Groningen, Department of Rehabilitation Medicine, Groningen, the Netherlands
[2]University of Groningen, University Medical Center Groningen, Center for Human Movement Sciences, Groningen, the Netherlands

ABSTRACT

Following an upper-limb amputation the muscles and tendons in the amputation stump are often rearranged by a surgical procedure. One of the purposes of this rearrangement is to shape the stump as to optimally support the prosthesis socket, and to create good control sites for a myoelectric prosthesis using direct control. For the transradial level the main wrist flexors and extensors are used for the latter purpose and the remaining muscles are mainly used for reshaping the stump. This is an interesting phenomenon from a motor control perspective and questions arise to how the control strategy of the neuromotor system changes after amputation when muscles and other tissues are rearranged and subsequently degenerate. Moreover, the feedback loop is heavily altered due to absence of a moving limb. This also appears to have an effect on the electromyogram (EMG) as demonstrated in several studies in which motion intent was classified using features of the EMG measured at the forearm. When comparing classification accuracy between able-bodied subjects and amputee subjects the accuracy was lower for the amputees. However, the relative accuracy between able-bodied participants and amputees is fairly consistent among a range of classification algorithms. Therefore, many studies recruit able-bodied subjects and extrapolate their results to the amputee population.

In this study we aim to investigate how transradial amputation influences the EMG in an effort to improve the transferability of results from able-bodied participants to amputee users. In our study protocol, we simultaneously measure the EMG at the forearm of both the unaffected and the affected side of transradial amputees. Participants will perform bimanual (phantom) movements in two different conditions. In the ‘restricted-hand condition’, the hand of the able side is restricted by a brace so the movement contractions become isometric. In the ‘free-hand condition’, the hand of the able side is not restricted. The purpose of restricting the able hand is to simulate the loss of hand movements while contracting wrist muscles and determine how this influences the EMG. We hypothesize that the EMG measured at the able-side in the ‘restricted-hand condition’ is more similar to the EMG at the affected side than it is in the ‘free-hand condition’. To quantify this, we use a pattern-recognition algorithm to classify the motion intent from both sides and analyse the resulting classification clusters using the separability index, repeatability index and the semi-principal axes as described in the literature.

RELATION BETWEEN CAPACITY AND PERFORMANCE IN PAEDIATRIC MYOELECTRIC PROSTHESIS USERS

[1]Helen Lindner, [1]Ayako Hiyoshi and [2]Liselotte Hermansson

[1]Örebro University
[2]Region Örebro County

BACKGROUND AND AIM

Myoelectric prostheses are often prescribed to children with upper limb reduction deficiency and training is given regularly by the prosthetic clinics. One goal of prosthesis fitting and training is to give the child a tool to assist when performing daily activities. Prosthetic fitting should be initiated at a young age but little is known whether the prosthetic skills training and recommendations for daily use of prostheses can ease the performance of daily activities. Measures of capacity and performance can help to determine if there is any gap between them that may restrict participation.

The aim was to explore the relationship between capacity scores obtained in a standardised clinical setting and proportional ease of performance in using the prosthesis to perform daily activities obtained from a real-life environment.

METHOD

During their clinic visits, pediatric prosthesis users (n=62, age 3 to 17) were asked to fill in a questionnaire, 'Prosthetic Upper Extremity Functional Index' (PUFI), where the child (or the parent if the child was under 6) rated the ease of performance in using the prosthesis to perform 26-38 daily activities. Then the child performed a bimanual activity and an occupational therapist from the clinic (n=6) assessed the child's capacity for prosthetic control with an observational based assessment, 'Assessment of Capacity for Myoelectric Control' (ACMC). In addition, the child or the parent was asked about the prosthetic wearing pattern. Sex and prosthetic side were recorded. Spearman correlation coefficient and Generalised linear model were used to examine the association between these measures.

RESULTS

A strong correlation (Spearman= 0.75) was found between the capacity scores and the ease of performance. In both unadjusted and adjusted models, capacity was significantly associated with proportional ease of performance. The adjusted model showed that, by 1 unit increase in the ACMC score, the ratio of proportional ease of performance increases by 45%. This implies that ACMC can be a predictor for ease of performance in real-life environment.

DISCUSSION & CONCLUSION

The ACMC as an independent variable was the strongest predictor variable for ease of performance. The results confirm earlier results suggesting a relationship between pattern of use and prosthetic skills. The conclusion is that wearing a myoelectric prosthesis every day facilitates learning of operation skill which, in turn, eases the performance of daily activities. Training for children with myoelectric prostheses should emphasize both establishment of wearing habits and practicing control skills during daily task performance.

A SELF-GRASPING HAND PROSTHESIS

Gerwin Smit, Bartjan Maat, Dick Plettenburg and Paul Breedveld

Delft University of Technology

BACKGROUND

This study presents an innovative approach for passive adjustable hand prostheses. Around a third of the upper limb amputees uses a passive prosthetic device, which can be a prosthetic hand or tool. In literature there has been very little attention for improvement of the function of passive adjustable (PA) hand prostheses.

GOAL

The goal of our study was to design a next generation adjustable prosthetic hand. This prosthetic hand must be able to grasp objects without the help of the sound hand, and without the need of a harness or batteries.

METHODS

An analysis of PA prostheses and relevant prosthetic characteristics was performed. We identified design requirements for a new and better PA prosthetic hand. The design of this new PA prosthesis mainly focused on two features; the grasping mechanism and the locking mechanism. For both features a function analysis was performed. Different working principles were designed and tested. A final prototype was designed, built and evaluated.

RESULTS

We designed an innovative passive prosthetic hand, the Delft Auto-grasping Hand (DAH). This hand has articulating fingers and can perform the hook grip, power grip and pinch grip. The gripping function is controlled indirectly by pushing an object to the hand, or directly by pushing the prosthetic thumb against a fixed object. The grip force is proportional to the applied push force. By releasing the push force, the grip force is locked and the object is being held. In order to release the object, a button has to be pushed after which the object can be released by pushing the object slightly into the hand. The DAH has a mass of only 130 grams. In an evaluation the DAH was compared with a conventional PA prosthesis. Activities were performed 11 % faster and required less user effort with the DAH. During the activities, the grasping function of the DAH was used 54% more often.

CONCLUSION

This study presents a next generation passive adjustable prosthetic hand, the Delft Auto-grasping Hand (DAH). The hand can grasp objects without the help of the sound hand. The DAH is the first PA prosthetic hand which has articulating fingers and can perform the hook grip, power grip and pinch grip. The evaluation showed that the DAH has a good grasping functionality and is easy to control. This innovative prosthetic hand offers an attractive alternative to current passive prosthesis, and possibly even to active prostheses.

DESIGN AND EVALUATION OF A NOVEL SENSORY-MOTOR TRANSHUMERAL PROSTHETIC SOCKET: A CASE STUDY

[1]Jonathon Schofield, [2]Michael Stobbe, [1]Jason Carey, [3]Paul Marasco and [1]Jacqueline Hebert

[1]University of Alberta
[2]Glenrose Rehabilitation Hospital
[3]Cleveland Clinic

ABSTRACT

This work describes a novel myoelectric transhumeral prosthetic socket designed to integrate a custom vibration tactor with improved fit, suspension, and secure electrode contact. A quantitative analysis was performed to evaluate the impact of the novel socket on the interface pressures between the socket and residual limb (RL).

A fundamental challenge in the implementation of advanced upper limb prostheses is the lack of sensory input. In response, haptic systems have been developed; however, practical barriers still exist in translating these systems beyond the benchtop into functional wearable prostheses. Most non-invasive systems employ mechanical devices (tactors) that stimulate strategic locations on the user's RL. A primary challenge lays in the development of prosthetic sockets allowing tactors access to the RL, while maintaining or improving socket fit. A well-fit prosthetic socket will secure the prosthesis to the limb ensuring suspension and security. Yet even well-fit traditional sockets are prone to slip during normal use. In myoelectric systems, this can create a loss of electrode placement resulting in poor or inconsistent control of the components.

Our approach integrated a previously developed tactor into a transhumeral socket, such that a predetermined distal anterior region of the participant's RL could be stimulated. A ¾" diameter window was created allowing the tactor access to the limb. The corresponding region on the participant's prosthetic liner was thinned. A flat build-up was added between the socket and prosthetic elbow providing a mounting surface. Custom brackets allowed attachment of Velcro strapping, providing adjustable affixment of the tactor. To ensure socket fit and electrode contact, a BOA Lace (Denver, USA) and an electrically conductive panel system were implemented. This system provided adjustable compression of strategic areas within the socket, with the panels also serving to make contact with the electrodes in the prosthetic liner. Therefore, tensioning the BOA system ensured firm electrode contact and distribution of pressure, while providing flexibility in the event of socket slip.

To evaluate the impact on socket fit, RL-socket contact pressures were captured using a Tekscan VersaTek system (Boston, USA). Relative pressure magnitudes and corresponding anatomical locations were compared across the novel socket and the participant's well-fit body powered prosthetic socket. Results highlighted reduced maximum pressure magnitudes spread more evenly across the RL while wearing the novel socket.

This work presents a unique solution to practical integration challenges associated with the development of functional sensate prostheses, with further applicability to myoelectric socket design in general.

Myoelectric Prosthesis Control: Does Augmented Feedback Improve Internal Model Strength and Performance?

Ahmed W. Shehata, Erik J. Scheme, and Jonathon W. Sensinger

Department of Electrical and Computer Engineering, Institute of Biomedical Engineering, University of New Brunswick, Fredericton, New Brunswick, Canada.

ABSTRACT

Amputees lack some of the sensory information that able-bodied persons incorporate in the use of their limbs; as a result, the control of myoelectric powered prostheses requires constant visual attention and a high level of concentration, which may lead to poor performance. Recent advances in signal processing and sensory technology have enabled the development of various methods of sensory feedback, including auditory, vibrotactile and electrotactile, which may be used to augment the feedback provided to prosthesis users. Researchers have explored the advantages of this augmented feedback by looking at the short-term performance results, but have not explored its effect on the development of the user's internal model, which affects the long-term performance. In this work, we investigate the notion that some controllers provide better short-term performance at the expense of providing inadequate feedback to develop a strong internal model, whereas other controllers may provide adequate feedback, but at the expense of more noisy control signals. We hypothesize that augmented feedback may be used to mitigate this tradeoff, ultimately improving short and long-term control. Using psychophysical assessment tools, we measured the internal models developed for three myoelectric controllers: 1) raw control with raw feedback (RCRF), such as a regression, 2) filtered control with filtered feedback (FCFF), such as a classifier, and 3) filtered control with audio augmented feedback (FCAF), such as a classifier control with augmented regression feedback. We assessed the short-term performance of these three control interfaces using a multi degree-of-freedom constrained-time target acquisition task. Results obtained from 30 able-bodied subjects showed that the FCAF control strategy enabled the development of a stronger internal model than FCFF, better accuracy and path efficiency than RCRF. These results support our hypothesis that the use of augmented feedback control strategies may improve both short-term and long-term performance.

JOINT-BASED VELOCITY FEEDBACK IMPROVES MYOELECTRIC PROSTHESIS PERFORMANCE

Eric J. Earley[1,2], Kyle J. Kaveny[2], Reva E. Johnson[3], Levi J. Hargrove[1,2,4], Jon W. Sensinger[5]

[1]*Department of Biomedical Engineering, Northwestern University, Chicago, IL USA*

[2]*Center for Bionic Medicine, Shirley Ryan AbilityLab, Chicago, IL USA*

[3]*Department of Mechanical Engineering, Valparaiso University, Valparaiso, IN USA*

[4]*Department of Physical Medicine & Rehabilitation, Northwestern University, Chicago, IL USA*

[5]*Institute of Biomedical Engineering, University of New Brunswick, Fredericton, NB Canada*

BACKGROUND

Those with upper-limb amputations have reduced sensory feedback, and this likely contributes to difficulties in performing daily activities [1]. Many attempts have been made to improve performance by providing sensory substitution, but few have succeeded with visual feedback present [2]. Research in computational motor control proposes three criteria for augmented feedback to be most useful. First, the feedback should provide information not available to other senses, notably vision [3]. Second, the feedback should have low uncertainty compared to the control of the task [4]. Third, feedback should provide information in the most uncertain reference frame (which, for EMG control, tends to be a local reference frame) [5]. These criteria suggest that a local, joint-based velocity feedback paradigm will improve prosthetic arm control, even for those with unaffected vision.

The aim of this study was to determine if local joint-based velocity feedback improves performance, even with vision present, during control of a 2 degree of freedom (DOF) myoelectric interface.

METHOD

Ten able-bodied subjects participated in the study, which was approved by our local ethics board. After providing informed consent, subjects controlled a myoelectric interface consisting of a virtual shoulder and elbow and were asked to perform time-constrained center-out reaches, arriving at the target within 1.5 seconds. Subjects completed one session with no audio feedback, and one session with audio feedback provided, where amplitude corresponded to joint speed, with a different frequency for each joint. After subjects were familiarized with the task, the simulated dynamics were perturbed by reducing the damping coefficient of the joints. We measured the increase in reaching error and average movement speed post-perturbation, and during reaches to different targets testing generalizability, and modeled the adaptation to these new system dynamics as an exponential decay function.

RESULTS

Subjects experienced a smaller increase in both reach errors and average speed immediately following the dynamic perturbation with audio feedback present. Though reaching errors were within baseline levels during the first generalization trial, speed increased by a smaller margin with audio feedback present.

DISCUSSION

These results suggest that local joint-based velocity feedback helped users recognize changed system dynamics and allow them to adapt faster to these new dynamics, even with vision feedback present.

ACKNOWLEDGEMENTS

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KINEMATIC COMPARISON OF BODY-POWERED AND MOELECTRIC PROSTHESES IN USERS WITH TRANSRADIAL AMPUTATIONS

Graci Finco[1], Kellie Keener[2], William Amonette[2], Thomas Krouskop[1]

[1]Department of Orthotics and Prosthetics, Baylor College of Medicine;

[2]Department of Clinical, Health, and Applied Sciences, University of Houston – Clear Lake

ABSTRACT

Objective: The purpose of this study was to quantify differences in shoulder, elbow, and wrist range of motion between myoelectric and body powered prosthesis users during three simulated ADLs. ROM of the involved limb and the sound limb were also compared. It was hypothesized that amputees utilizing a myoelectric prosthesis would exhibit less ROM for a given task compared to that of body powered users, and that the prosthetic limb would require greater ROM in all joints than the sound limb to accomplish a task.

Methods: Three subjects participated in this project. One subject used a myoelectric prosthesis, one used a body-powered prosthesis, and one subject used both. Volunteers performed three simulated ADL's with their sound and prosthetic limb: object transfer, drinking from a cup, and hair combing. Three dimensional kinematic data were collected using an eight camera passive optical motion capture system (Vicon, Denver, CO). A 23-marker model was used for data collection. During processing, the markers were manually labeled and gaps were filled using a Woltring algorithm. Upper-limb joint kinematics were modeled to quantify shoulder, elbow, and wrist ROM on the amputated and sound limb during the three selected tasks. Each task was repeated three times, and the middle trial of each task was used to minimize learning effect.

Results: Using the prosthesis did not necessarily require greater ROM in all joints than the sound side in all subjects. All subjects exhibited different movement strategies; body-powered subject 1 typically used the least ROM and body-powered subject 2 used the greatest ROM to accomplish tasks despite identical componentry. The subject that used both a myoelectric and body-powered prosthesis tended to use greater ROM with the myoelectric prosthesis to accomplish tasks.

Conclusion: This pilot study with a small sample size provided unexpected results and highlighted the importance of socket comfort, formal prosthetic training, and choice of components as critical factors prosthetists can control that affect ROM in users of transradial prostheses.

INTRODUCTION

In an early study centered around the time of initial myoelectric clinical acceptance in 1983, Stein and Walley compared myoelectric and body-powered prostheses through a series of standardized tasks. Users of myoelectric prostheses scored higher in tests of functional range of motion, and were able to carry out the tasks with less compensatory movements. They also found that body-powered users took 2.5 times longer and myoelectric users took 5 times longer to complete the tasks as compared to their sound side. This was a primary study and used as a basis for later research [1].

Carey et al found that users of transradial myoelectric prostheses had decreased humeral flexion and increased elbow flexion when compared to able-bodied individuals while drinking from a cup [2]. Additionally, Metzer et al discovered users of transradial prostheses had larger shoulder and elbow path distances than in able-bodied subjects while performing ADLs [3].

The prosthesis best suited for the amputee's needs depends on control, function, feedback, cosmesis, and rejection according to a systematic literature review by Carey et all in 2015. This review focused on differences between myoelectric and body-powered prostheses. They concluded that current evidence is insufficient to show functional differences between myoelectric and body-powered prostheses [4].

This study will attempt to address this lack of evidence in functional differences between myoelectric and body-powered prostheses by determining functional differences between motion envelopes of myoelectric and body-powered prostheses. This data will be compared to provide insight into compensation strategies within subjects between their sound and amputated side as well as between users of myoelectric and body-powered prostheses.

METHODS

Participants

Unilateral transradial amputees with residual limb length ranging from extremely short to a wrist disarticulation were included in this study. Subjects were users of a myoelectric or body powered prosthesis for at least six months or more, and were screened to ensure they had the range of motion required to operate their prosthesis. Subjects under the age of eighteen were excluded from this study as well as subjects with a neurological or musculoskeletal pathology that impairs arm motor control. Additionally, subjects that had weakness in their forearm defined as a manual muscle testing score of 3+ or less on their amputated arm were excluded from the study.

Experimental Protocol

Once the subjects arrived at UH-Clearlake's facility they signed a consent form and were tested for strength and range of motion to verify that they qualify for the study. Then, they answered a brief survey inquiring about their time since amputation, cause of amputation, years of prosthetic experience, socket comfort score, and completed the Upper Extremity Function section of the OPUS (Orthotics and Prosthetics Users Survey) outcome measure questionnaire. Reflective markers were placed on the subjects and motion analysis was conducted after the questionnaires were completed.

Participants were asked to execute one goal-oriented task listed on the Southhampton Hand Assessment Procedure (SHAP), and two goal-oriented tasks from the Upper Extremity Function section of the OPUS. All tasks were completed while seated in a chair at a table of standard height at 18.25 inches and 28 inches respectively.

1. Lifting and transferring a weighted object- a standard mason jar was closed with a lid. The subject must lift the jar with the prosthetic side, transfer it one foot as specified with tape on the table over a two inch barrier, and set the jar down on the tape.

2. Drinking from a cup-an empty standard sized Solo cup was used. They grasped the empty cup without crushing it and lifted it up to their mouth as if to drink.

3. Hair combing- a comb was raised from the top of their head to the back of their head above their hair.

All of these tasks required grasping an object that can be done with any terminal device in the same neutral position. Participants were asked to complete task requirements as accurately and quickly as possible. Each subject started and ended each task with their arms resting on the table in a neutral position. They were permitted practice to familiarize themselves with the task prior to data collection. Each task was repeated three times with their amputated side first, then repeated three times with their sound side.

Data collection and analysis

Kinematic data collection of upper extremity joint angles, patterns, and positions were accomplished using an eight camera passive optical Vicon 3D Motion capture system. Approximately 23 reflective markers were adhered to the skin or clothing using double-sided tape in a standard configuration consistently placed on subjects by the same individual. The three dimensional coordinates of marker data were used to reconstruct joint angles, calculating kinematic parameters using an upper-body model plug-in with the NEXUS (Vicon Nexus, Denver, CO) software.

The anatomical locations of the markers used in this analysis include: C7 spinous process, T10 spinous process, right scapula, sternum, bilateral acromion processes, bilateral triceps, bilateral biceps, medial and lateral epicondyles, bilateral forearms, radial and ulnar styloids, and bilateral third MCP joint. The epicodyle, forearm, styloid, and third MCP joint markers on the prosthetic side were placed at the relative position of the anatomical locations on the subject's sound side.

Statistical analysis

A 23-marker model was used for data collection. During processing, the markers were manually labeled and gaps were filled using a Woltring algorithm with a minimum gap length of 5 consecutive points. Upper-limb joint kinematics were modeled to quantify shoulder, elbow, and wrist ROM on the amputated and sound limb during the three selected tasks. Joint kinematics were taken of the middle trial of each task to minimize learning effect. The only cases in which the middle trial was not analyzed included if the task was not completed or an anomaly occurred that prevented processing of the task.

Joint kinematics were exported to Excel and absolute maximum and minimum joint range of motions were found using Excel formulas for each task. The absolute maximum and minimum joint range of motions were subtracted to find the total change in joint angles in degrees. This was done for the sagittal, coronal, and transverse planes of the shoulder, the sagittal and coronal planes of the wrist, and the sagittal and the sagittal plane for the elbow. Angular velocity is still being calculated.

RESULTS

Three subjects with traumatic amputations participated in this study. One subject used a myoelectric prosthesis, one used a body-powered prosthesis, and one used both a myoelectric and body-powered prosthesis. The subject using a myoelectric prosthesis was right hand dominant and her amputated side was her left side. The other two subjects were right hand dominant and became left hand dominant after their right side was amputated. The subject that was only tested using a body-powered prosthesis also has a partial hand amputation on his left side.

Experience using a prosthesis ranged from 1.5 to 19 years. Socket comfort scores averaged a 7.5 rating out of a 10 point scale. Average age, height, and weight were 57.3 years, 5'9, and 180lbs. Average time of prosthetic wear was 12 hours. Residual limb length ranged from 4 inches to 8 inches. A comprehensive chart of subject information can be found in Table 3, but notably both myoelectric prostheses had wrist rotators and both body-powered prostheses had figure of 9 harnessing, friction wrists, and 5XA hook terminal devices. One subject using a myoelectric prosthesis used a hand as the terminal device, while the other used an electronic hook as the terminal device.

Please refer to Figures 1-4 and Tables 1-3 to reference numerical results. Results indicated the prosthetic side did not necessarily require greater ROM than the sound side in all subjects and in all joints. The results indicated an even split between whether the prosthetic side or sound side used more ROM. Both subjects that used body-powered prostheses exhibited different movement strategies; body-powered subject 1 typically used the least ROM and body-powered subject 2 used the greatest ROM to accomplish tasks. The subject that used both a myoelectric and body-powered prosthesis tended to use greater ROM with his myoelectric prosthesis to accomplish tasks.

DISCUSSION

Myoelectric subject 1 (Myo 1) and body-powered subject 2 (BP 2) used greater ROM in all planes in all tasks than the subject that was tested using both his myoelectric and body-powered prosthesis (Myo 2 and BP 1). This could have been due to differences in time since amputation. Myo 1 had their amputation a year and a half prior to testing and BP 2 had their amputation 2 years prior to amputation. Meanwhile, the subject listed as Myo 2 and BP 1 had more experience using both of his prostheses. This increased experience may mean increased proficiency performing ADLs, which may have enabled the subject to use less ROM to accomplish tasks.

The differences in ROM between the two body-powered subjects may be explained by time since amputation as previously discussed. Additionally, two other factors may have affected this difference- the amount of formal training the subject had with the prosthesis and their self-reported Socket Comfort Scores. BP 2 reported extensive Occupational Therapy training with the prosthesis, while BP 1 had minimal Occupational Therapy training with the prosthesis. This lack of formal training may have contributed to why BP 2 had increased ROM in all tasks compared to BP 1.

Furthermore, BP 1 had a self-reported Socket Comfort Score of 8 out of 10 when asked how comfortable he found his prosthetic socket, while BP 2 rated his prosthetic socket at 6 out of 10. Since BP 1 found their socket comfortable and BP 2 stated they found their socket uncomfortable, this may have contributed to the difference in ROM in the two subjects. BP 2 may have used more ROM in each task to accommodate for an ill-fitting socket.

Myo 1 used more ROM than Myo 2, which may have been due to the subject's time since amputation as previously discussed. However, hand dominance and prosthetic components may have also contributed to the differences in ROM. Myo 1 remained right hand dominant after the amputation, whereas Myo 2 had to switch hand dominance from right to left after the amputation. Perhaps the biggest reason for this difference in ROM was due to componentry. Myo 1 used a hand as the terminal device of the prosthesis, while Myo 2 used an electric hook (ETD) as their terminal device. The anatomical hand cover on Myo 1's terminal device results in decreased line of sight, which may explain the increased ROM used to grasp and manipulate objects.

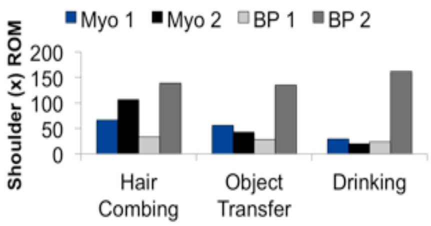
There seemed to be no correlation between the self-reported level of difficulty from the OPUS survey and the ROM measured from the subject using motion analysis. A future study could also include movement of the furthest point, or finger marker, from the torso, or T10 marker, as this measurement would be of excursion the body captures necessary to operate the prosthesis, rather than the total ROM of each joint.

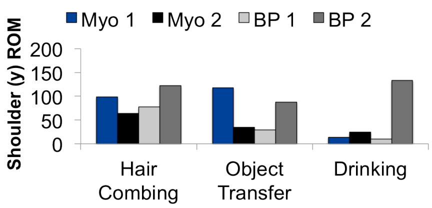
CONCLUSION

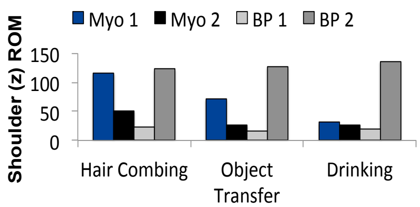
This case series analysis does not support the commonly held clinical opinion that users of body-powered prostheses typically use more ROM than users of myoelectric prostheses to accomplish ADLs. Additionally, the prosthetic side did not use substantially more ROM than that of the sound side to complete the three tasks. The small sample size in this pilot study produced unexpected results that indicate common held clinical opinions may need to be reexamined.

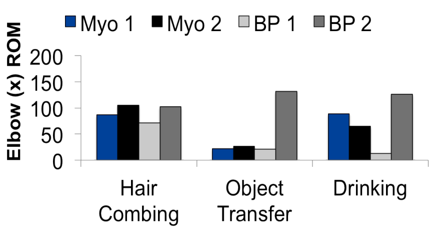
This pilot study highlighted the importance of socket comfort, formal prosthetic training, and choice of components as critical factors prosthetists can control that affect ROM in users of transradial prostheses. Hand dominance and time since amputation should also be considerations when deciding on a type of prosthesis and components for a patient to avoid limiting their ROM. Furthermore, the results from this study emphasize the importance of using motion capture to investigate ROM in upper-limb prosthesis users as well as outcome measures for a more accurate analysis.

FIGURES AND TABLES









Figures 1-4: Degrees of shoulder flexion/extension (x), ab/adduction (y), int/external rotation (z), and elbow flex/ext (x) ROM during tasks.

Table 1: Comparison of shoulder, elbow, and wrist ROM in degrees between prosthetic (PS) and sound (SS) limb during hair combing.

	Myo 1		Myo 2		BP 1		BP 2	
	PS	SS	PS	SS	PS	SS	PS	SS
Sho (x)	66.3	76.5	106.6	57.9	32.9	57.9	138.6	118.8
Sho (y)	98.3	51.1	63.6	61.2	77.5	61.2	122.7	91.5
Sho (z)	115.9	117.4	51.2	68.0	22.3	68.0	122.9	89.5
Elb (x)	86.4	87.0	105.1	67.3	71.4	67.3	102.0	106.8
Wri (x)	58.0	83.4	32.0	5.3	135.4	5.3	65.6	58.2
Wri (y)	19.4	73.8	23.5	3.0	44.0	3.0	88.6	31.0

Table 2: Comparison of shoulder, elbow, and wrist ROM in degrees between prosthetic (PS) and sound (SS) limb during object transfer.

	Myo 1		Myo 2		BP 1		BP 2	
	PS	SS	PS	SS	PS	SS	PS	SS
Sho (x)	55.8	115.1	42.5	40.9	27.9	40.9	135.2	54.6
Sho (y)	118.1	116.2	34.8	34.4	29.1	34.4	87.6	32.8
Sho (z)	71.1	109.5	27.3	58.2	15.9	58.2	127.2	61.0
Elb (x)	21.9	97.1	26.8	25.9	21.6	25.9	131.2	52.8
Wri (x)	2.9	38.3	2.2	26.4	4.7	26.4	46.6	9.0
Wri (y)	2.3	10.2	1.2	15.2	2.8	15.2	4.8	5.9

Table 3: Comparison of shoulder, elbow, and wrist ROM in degrees between prosthetic (PS) and sound (SS) limb during drinking.

	Myo 1		Myo 2		BP 1		BP 2	
	PS	SS	PS	SS	PS	SS	PS	SS
Sho (x)	29.5	21.4	20.1	22.2	23.3	22.2	161.8	93.6
Sho (y)	13.5	39.8	24.2	15.8	10.3	12.9	133.4	80.0
Sho (z)	31.0	18.0	26.4	17.5	20.0	17.5	136.4	150.3
Elb (x)	88.4	113.9	64.6	81.2	13.1	81.2	126.0	144.1
Wri (x)	11.9	6.3	3.8	21.7	10.2	21.7	77.8	65.1
Wri (y)	9.8	15.2	3.7	10.1	5.3	10.1	72.8	45.1

EQUATIONS

For purposes of this study only the total change in ROM in degrees is compared. Angular velocity for all data is still being calculated.

absolute maximum- absolute minimum ROM total

(change in ROM) / .01 Angular vel.

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INTERROGATING THE FUNCTIONAL INTERPRETATION OF JOINT MOVEMENT ILLUSIONS USING INTENTIONAL BINDING

Courtney Shell and Paul Marasco

Laboratory for Bionic Integration, Lerner Research Institute, Cleveland Clinic

ABSTRACT

Sensation of joint movement provided through a vibration-induced illusion has potential use in restoring lost kinesthetic sensations, such as those caused by amputation. In order to be usefully employed, the way in which sensations provided by the illusion are incorporated into the body's internal model for motor control must be explored. While literature suggests that vibration-induced illusion of a joint movement is generated by providing vibration to the antagonist muscle (e.g., elbow flexion illusion induced by vibrating the triceps), perception of limb movement appears to be more complex as vibration of a given muscle in targeted reinnervation amputees generates an illusion of joint movement associated with contraction of the vibrated muscle. To explore how vibration-induced illusion of joint movement is interpreted by the body's internal model, we investigated perceived compression of time (intentional binding) between an auditory signal and completion of a participant-controlled virtual arm movement paired to the movement illusion. In this paradigm, when conditions are more natural subjects experience compression of the time interval between an action and results of the action. Thus, the movement of a virtual arm shown to the subject that most closely matches the internal model's interpretation of the vibration-induced illusion can be identified.

SERVICE MEMBERS AND VETERANS WITH TRANSHUMERAL OSSEOINTEGRATION: INITIAL REHABILITATION EXPERIENCES FROM THE DOD OI PROGRAM AT WRNMMC

[1]Annemarie Orr, [2]Michelle Nordstrom, [1]Mark Beachler, [1]Josef Butkus, [2]Brad Hendershot, [1]Barri Schnall, [1]David Laufer, [1]Stanley Breuer, [3]Paul Pasquina, [4]Benjamin Potter, [4]Jonathan Forsberg and [5]Christopher Dearth

[1]Walter Reed National Military Medical Center
[2]The Henry M. Jackson Foundation
[3]Department of Rehabilitation Medicine, USUHS
[4]Department of Orthopedics, WRNMMC
[5]DOD-VA Extremity Trauma and Amputation Center of Excellence

BACKGROUND

Since September 11, 2001, the Armed forces sustained a total of 1,706 combat amputations in which 296 had upper extremity involvement. In 2010 it was identified that 22% of the Operation Iraqi Freedom/Operation Enduring Freedom Veterans with unilateral upper-limb amputations have completely abandoned their prosthetic devices and the percent of Vietnam Veterans who abandoned their prosthesis was 30%. Prosthetic abandonment is due to many factors including pain, weight, skin breakdown, and lack of consistent function.

Osseointegration has been performed internationally for facial injuries, hearing aids, finger joints, and limb prostheses over the last two decades. Initial procedures in United States were performed for lower limb amputations beginning in 2015 with the first FDA approved devices becoming available in 2016.

Walter Reed National Military Medical Center is working to reduce the rate of abandonment through the implementation with Osseointegration (OI); using a direct skeletal attachment technique developed by P-I Brånemark from Gothenberg, Sweden called Osseointegrated Prostheses for the Rehabilitation of Amputees (O.P.R.A.). Initial enrollees in the clinical trial for transhumeral amputees have had long standing issues with prosthetic functionality leading to abandonment or limited use, but desire to use their prosthetic device. This abstract is intended to describe the rehabilitation protocol, rehabilitation timeline, and lessons learned from the first three upper limb OI participants.

METHODS

The rehabilitation protocol between the two surgeries includes wound care, range of motion (ROM), and strengthening with a clinical ROM evaluation conducted every two to three weeks. The goal between the first and second surgeries is to maintain ROM and strength. Three to four weeks after the second surgery a training prosthesis is incorporated into the treatment plan to gradually increase weight tolerance. Assessments are performed pre and post-operatively over a 24 month timeframe. Evaluations include Goniometric and Biomechanical ROM measurements, ACMC, UNB, Box & Blocks, and pinch pins.

RESULTS

Preliminary results of the first three participants self-report using the Visual Analog Pain Scale, DASH, and PROMIS Questionnaires minimal discomfort in between surgeries. ROM and strength were regained following their home exercise program (HEP). No clinical setbacks (infection, surgical complications, or excessive pain) impacted the rehabilitation progress of the initial participants. All three reverted back to their previous prosthetic use between Stage 1 and Stage 2 surgeries.

CONCLUSION

The OI procedure has given patients the opportunity to explore new avenues, improve prosthetic functioning, and quality of life.

EMBODIMENT OF BI-DIRECTIONALLY INTEGRATED PROSTHETIC LIMBS

[1]Courtney Shell, [1]Rafael Granja-Vazquez, [2]Zachary Thumser, [1]Dylan Beckler and [1]Paul Marasco

[1]Laboratory for Bionic Integration, Lerner Research Institute, Cleveland Clinic
[2]Louis-Stokes Cleveland Department of Veterans Affairs Medical Center

ABSTRACT

The sense that a prosthetic hand is something other than the body, that it is a tool, prevents instinctive engagement with the device by amputees. We have built myoelectrically-controlled, battery-powered prosthetic limb systems with robotic sensation for home use that provide tactile feedback when the prosthetic fingers contact objects. These systems are currently employed in a take-home trial with amputee participants that have a biological neural machine interface (targeted reinnervation). We identified locations on the reinnervated skin of three participants that correspond to a feeling of touch on their missing fingers and matched them to sensors integrated with the terminal device digits (strain gages in D1-D3 and force sensitive resistors on D4 and D5). When sensors on the prosthesis detect contact, touch robots mounted above these locations press on the reinnervated skin to generate a feeling of proportional pressure that is projected to the appropriate missing fingertip. During baseline testing at the start of the ten month study period, we investigated whether tactile feedback during a series of psychophysical tests would induce a sense of ownership (i.e., embodiment) of the robotic prosthetic hand. Subjects completed questionnaires indicating the degree to which they agreed with nine different statements (three embodiment-related and six control). Two users showed greater embodiment of the prosthetic hand when tactile feedback was provided. The third user showed a slight trend toward embodiment when using his prosthesis both with and without tactile feedback provided by the touch robots. Providing a sense of touch to prosthesis users through a bi-directionally integrated limb encourages embodiment of the prosthetic hand so that it is interpreted as being part of the amputee’s body.

MEASURING USER EXPERIENCE OF A SENSORY ENABLED UPPER LIMB PROSTHESIS

[1]Linda Resnik, [2]Emily Graczyk and [2]Dustin Tyler

[1]*Providence VA Medical Center*
[2]*Case Western Reserve*

BACKGROUND

New technologies that restore sensory feedback to upper limb prosthesis users have the potential to greatly improve quality of life. One such technology is the sensory restoration systems (SRS) developed at Case Western Reserve University. Measuring the impact of SRS is challenging, given that existing measures do not quantify likely psychological impacts of SRS.

PURPOSE

To describe the development of a multi-dimensional subjective experience scale that is responsive to change associated with use of an SRS.

METHODS

Content development

Measure content was identified through informal conversations with two subjects implanted with an SRS, discussion with subject matter experts and literature review. Preliminary item banks were drafted and reviewed by measurement workgroup members. Items were refined based on feedback. Subscales were created for: self-efficacy of prosthesis use, prosthesis embodiment, body image, prosthesis efficiency and social touch.

Patient Experience Measure

Items are graded using a 5 point Likert Scale (strongly disagree to strongly agree). The self-efficacy subscale asks subjects to rate confidence using the prosthesis to complete 7 items which are typically challenging for prosthesis users. The Embodiment subscale consists of 8 items that ask about prosthetic embodiment (e.g. the prosthesis is a part of me) The 9-item Body Image subscale asks about impact of the prosthesis on self-image (e.g. when I remove my prosthesis I feel more confident). The 3-item Prosthesis Efficiency scale includes items relating to speed and focus required to use the prosthesis. Finally, the social touch subscale consists of 11 items pertaining to prosthesis use in social interactions.

Data collection

Two subjects with implanted SRS participated in a home study. During the intervention stage, each wore an experimental hand system with embedded sensors and received nerve stimulation. During the Pre-test and Post-test stages, subjects wore the experimental hand system, without stimulation. At the end of each stage, subjects completed the Patient Experience Scale.

Data analysis

Item scores for each subscale were averaged. Descriptive analyses were conducted by subject and stage.

RESULTS

Scores for self-efficacy, embodiment, efficiency and social touch subscales were higher for the sensory stimulation stage for both subjects. Scores for body image were highest for subject 1 at post-test and highest for subject 2 during sensory stimulation.

CONCLUSIONS/IMPLICATIONS

Findings provide preliminary evidence of the validity and responsiveness of the Patient Experience Scale, a unique measure designed to quantify impact of prosthetic sensory restoration. Data collection in additional subjects will enable examination of scale internal consistency.

SPATIAL FILTERING FOR ROBUSTNESS OF MYOELECTRIC CONTROL ON ELECTRODE SHIFT

Jiayuan He[1], Xinjun Sheng[2], Xiangyang Zhu[2], Ning Jiang[1*]

[1]*Department of Systems Design Engineering, University of Waterloo, Waterloo, Canada*

[2]*State Key Lab of Mechanical System and Vibration, Shanghai Jiao Tong University, Shanghai, China*

ABSTRACT

Electrode shift is one of the factors that degrade the myoelectric control performance. In this study, two spatial filters, Laplacian filter (LF) and circular average filter (CAF), were separately applied on four channels of surface EMG signals, and their respective effects on the classification performance with and without electrode shift were investigated. The results on a classification task of eleven hand and wrist movements showed that CAF could significantly decreased the error rates with electrode shift, while LF significantly increased the error rates. The outcome of this study would benefit the design of the electrodes and increased the robustness of the PR-based myoelectric control.

INTRODUCTION

Pattern recognition (PR) algorithm could provide intuitive and dexterous control of multi-functional myoelectric prostheses for the users with motor deficits [1]. For the application of the PR-based control scheme on the commercial prostheses, the key problem is its robustness against the disturbances in activities of daily life (ADL) [2-5], such as arm position movement, muscle fatigue, electrode-skin contact condition change, electrode position change, etc. Among these factors, electrode position change is inevitable between donning and doffing, and would cause dramatic performance decrease in system control [6]. As such, attentions were received and multiple methods have been proposed to reduce the effects of electrode shift, ranging from the training strategy, electrode configuration to the feature extraction and classifier selection [6-8].

As the electrode position change led to the spatial changes of EMG signals, the application of the appropriate spatial filters could potentially improve the classification performance under electrode shift. Recently, some advanced spatial filters were applied on the high density (HD) electrode gird and low classification errors were achieved when electrodes shifted [9-10]. Considering the problem of practical use of the HD electrode grid in current socket systems, this study focused on the simple spatial filter operators with a small number of electrodes. Two spatial filters, Laplacian filter (LF) and circular average filter (CAF), were investigated in this study and their effects on classification performance with and without electrode shift were investigated.

METHODS

<u>Data Collection</u>

Nine able-bodied subjects (all males, from 20 to 30 years old) participated in the experiment. The informed consent was obtained before the experiment and the procedures were in accordance with the Declaration of Helsinki.

Eleven classes were investigated in this study, which were hand open, hand close, wrist flexion, wrist extension, radial flexion, ulnar flexion, pronation, supination, fine pinch, lateral prehension and rest. One run was defined as one repetition of these eleven classes, and each contraction lasted 5 s. A total of 16 runs were performed for one subject. The rest time was 5 s between two consecutive contractions and 30 s between two consecutive runs.

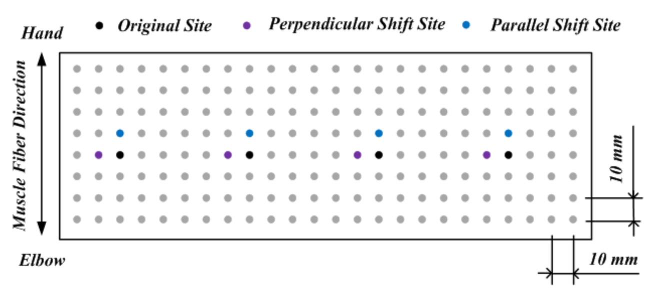

Figure 1: The original electrode site and its corresponding shift site.

A high density (HD) grid with 192 monopolar electrodes were used to collect EMG signals from the forearm. The inter-electrode distance was 10 mm and the grid was approximately 30 mm distal to the elbow crease. The signals were amplified with a commercial system (EMG USB2+, OT Bioelettronica, Italy) and sampled at 2048 Hz. As the current practical socket design would not allow for hundreds or even tens of channels, only four channels, evenly spaced around the forearm, were considered in this study (Fig. 1). Three cases were investigated, which were no shift, 10 mm shift parallel to muscle fibres, and 10 mm shift perpendicular to muscle fibres. The 10 mm shift distance was chosen for it was more likely in the daily life use situations [7].

Signal Processing

Two spatial filters common used in the image processing, Laplaician filter (LF) and circular averaging filter (CAF), were investigated in this study. LF is a high-pass spatial filter while CAF is a low-pass filter [11]. The operator of each filter is a three dimensional matrix (Table 1). The baseline (BL) is defined as the classification performance with four channels without filter. The signals of one channel are filtered by weighted summation of its own and neighbouring recordings. Suppose the filter operator matrix is S[i,j], the value of EMG signal from the channel located at row m, column n is E[m,n], the filtered signal F[m,n] is

$$F[m,n] = \sum_{i=1}^{3}\sum_{j=1}^{3} E[m-2+i, n-2+j] \times S[i,j]$$

The raw signals were segmented into 200 ms windows, with an overlap of 150 ms. Four time domain features, *i.e.* mean absolute value, zeros crossings, slope sign changes, waveform length [12], were extracted from each window. The classifier was linear discriminant analysis (LDA) [13] and the fold of cross validation was two.

Table 1: Spatial Filter Operator

Name	Matrix
Laplacian Filter (LF)	$\begin{bmatrix} 0 & -1 & 0 \\ -1 & 4 & -1 \\ 0 & -1 & 0 \end{bmatrix}$
Circular Averaging Filter (CAF)	$\begin{bmatrix} 0 & 1 & 0 \\ 1 & 0 & 1 \\ 0 & 1 & 0 \end{bmatrix}$

Statistical Analysis

A two-way ANOVA was conducted on the classification errors to compare the methods and shift conditions. Focused ANOVA would be performed by fixing the levels of one factor when the interaction between two main factors was significant in the full model. When significance was detected for the main factors, Tukey comparison was performed. The significance level was 0.05.

RESULTS

The classification errors of all the three methods were increased with electrode shift (Fig. 2). In all scenarios (with or without shift), the error rates of CAF, was either the smallest or one of the smallest, while LF was always the worst. The results of the two-way ANOVA revealed that there were significant interactions between the factors of methods and shift conditions. The following focused ANOVA showed that when there was no shift, the performance of CAF and BL was significantly better than that of LF, while no significant difference was detected between BL and CAF. For both shifting scenarios, the performance of CAF was significantly better than that of LF and BL, and the performance of BL was significantly better than that of LF.

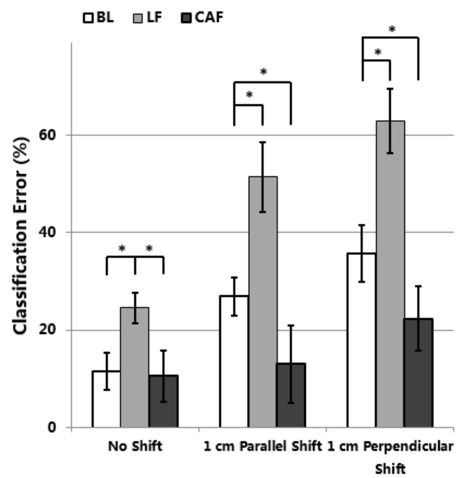

Figure 2: Classification errors of three methods, baseline (BL), Laplacian filter (LF) and circular averaging filter (CAF). The results are averaged across nine able-bodied subjects. Error bars represents the standard deviation. The star (*) represents that significant difference is detected between two corresponding methods.

DISCUSSION

This study investigated the effect of two spatial filter, LF and CAF, on the classification of eleven hand and wrist movements with and without electrode shift. Spatial filters were approved to be effective in improving myoelectric control performance [9-10]. The simplicity of the filters adopted in this study made them possible to be implemented in the real-world myoelectric prostheses control. It was observed that the classification errors after shift were greatly decreased by the application of CAF, while LF had the opposite effect (increasing error in all three cases investigated). This result is indeed expected: as CAF is a low-pass filter, it extracts information that is insensitive to the spatial variations. On the contrary, LF is high pass and extracts information sensitive to the spatial variations. As electrode shift caused changes in spatial domain, it was reasonable that the robustness of the system was increased by CAF, and decreased by LF. It was unexpected that the performance without shift was decreased by LF. The high sensitivity of LF to the noise might be the reason for this phenomenon, and it could be overcome by the combination of a low pass filter [11], such as Gaussian filter.

Only 4 channels were used in the calculation of the BL results, while 16 channels (12 neighbouring channels) were used in the calculation of the results for LF and CAF, which made the comparison biased to the advantages of the spatial filters. However, as the extra 12 channels were all located around the 4 channels with 10 mm distance, the signals they detected would be similar to each other. Therefore, there would be no big difference between the classification results they achieved.

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A NOVEL PASSIVE COMPLIANT WRIST WITH AUTOMATIC SWITCHABLE STIFFNESS

Federico Montagnani, Marco Controzzi, and Christian Cipriani
The BioRobotics Institute, Scuola Superiore Sant'Anna, V.le R. Piaggio 34 56025 Pontedera, Italy

ABSTRACT

State of art upper limb prostheses lack several degrees of freedom (DoF) and force amputees to compensate for them by changing the motion of their arms and body. Such movements often yield to articulation injuries and in general represent a discomfort; nonetheless these could be prevented by adding DoFs, for instance, to an articulated passive wrist. Available stiff or compliant wrists with passive flexion/extension and/or radial/ulnar deviation are sub-optimal solutions. Indeed, stiff wrists induce the individuals wearing them to perform exaggerated compensatory movements during the reaching phase while compliant wrists proved to be unpractical while manipulating heavy objects. Here we present the concept of a wrist capable of combining the benefits of both stiff and compliant wrists. It is provided with two switchable levels of passive compliance that are automatically selected depending on the grasp phase.

INTRODUCTION

The development of a natural hand prosthesis as a substitute of the biological limb, after amputation, is one of the most fascinating and open challenges in rehabilitation engineering. The limits that prevent the advent of next generation prostheses are well known and pertain to both the human machine interface and the physical features of the device. Among the latter, the most crucial is probably the lack of compact and reliable actuators with power densities similar to the human muscles. This deficit, combined with design trade-offs pertaining to desired performance, control inputs, prehension capabilities and anthropomorphism, implies that a hand prosthesis can perform a reduced set of movements only with respect to the natural counterpart [1].

The design of currently available prosthetic wrists represents a striking example of such a simplification. With its three degrees of freedom (DoFs), the natural wrist contributes to the execution of a grasping and manipulation task, by orienting the hand in space. However, in modern upper limb prostheses, such elegance is synthetized within a single DoF; myoelectric wrists are primitive, albeit useful, rotators that enable to pronate/supinate the hand [2].

Wrists with passive flexion/extension and/or radial/ulnar deviation were also demonstrated and made commercially available; these can be classified into *stiff* and *compliant* wrists. Stiff wrists enable the user to manually orient and lock the hand in a desired and firm position [3]. Compliant wrists can also be manually locked in a certain position, but when unlocked they exhibit an elastic behavior. Hence, as their name suggests, they allow for adaptation of the prosthesis during reaching and grasping, as well as other Activities of Daily Living (ADLs), e.g. bike-riding [4].

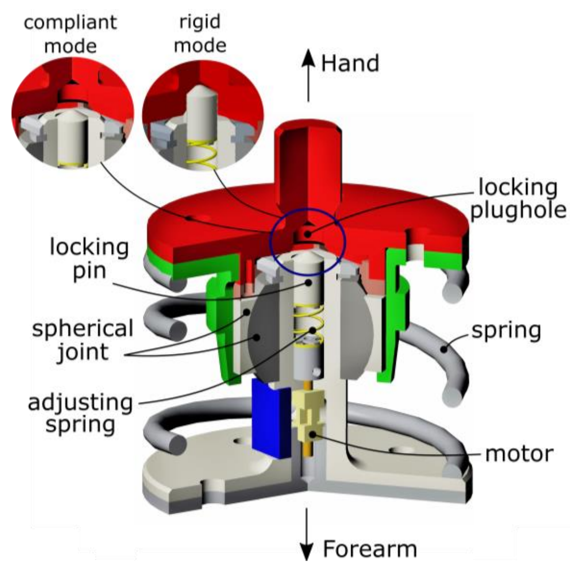

Fig. 1 Cross section of the wrist.

A number of studies compared the performance of stiff versus compliant wrists during ADLs [4]-[6]. All of them revealed improved functionality for most of the ADLs (in particular: bimanual tasks and tool manipulation) when using the compliant wrist, with the exception of those tasks which involved the manipulation of heavy objects; these tasks were performed better using a stiff wrist. These studies highlighted the limitations of both kinds of passive wrists. In particular, stiff wrists force the amputees to perform exaggerated compensatory movements during the reaching phase [7] (which are known to cause discomfort and secondary injuries in the long run [8]). Compliant wrists proved impractical while manipulating objects, particularly heavy ones.

In the light of these findings, we developed a concept of a novel compliant wrist with automatically selectable stiffness. The concept is aimed at combining the benefits of a compliant wrist (i.e. dexterity during the reaching phase of objects limiting compensatory movements) with the benefits of a rigid wrist (i.e. precision and safety while manipulating heavy objects).

SYSTEM ARCHITECTURE

The compliant wrist presents two automatically selectable levels of stiffness. The device (Fig. 1) consists of a spherical joint (range of motion ± 30 degrees) with its two sides connected by means of a compression spring. This spring is responsible for the compliant behavior of the wrist. To switch from *compliant mode* to *stiff mode*, the spherical joint can be locked using a linear actuation system based on a squiggle motor (New Scale Technologies, Inc., NY, USA) that drives a locking pin into a plughole. In compliant mode, the spherical joint is free to move under tangential forces applied to the prosthesis. Within this configuration, the bending of the spring is responsible for the elastic response of the wrist. When switched to *stiff mode* the locking pin is driven forward by means of the actuation mechanism and inserted in the plughole of the frame when the wrist is in its rest position (i.e. plughole centered with respect to the pin, Fig. 1). In the case of switching from compliant mode to stiff mode while the mechanism is not in this rest position, the adjusting spring ensures that the locking pin enters the plughole as soon as the two get aligned.

OPERATION OF THE WRIST

The wrist was designed to aid amputees in manipulation using a myoelectric prosthesis. The typical manipulation starts from a rest position and consists of reaching, grasping and holding phases (Fig. 2). During reaching, the arm transfers the hand towards the target and the hand is preshaped according to the dimensions of the manipulandum. The reaching phase ends with the enclosing of the manipulandum (grasping phase). The wrist is compliant during reaching to facilitate the positioning of the hand with respect to the manipulandum. This is obtained by pushing the hand against constraints in the environment (e.g. a vertical wall, a horizontal shelf, etc.) or the manipulandum itself (Fig. 2). Once the hand encloses the manipulandum, the wrist automatically switches from compliant to stiff mode in order to make the amputee able to safely manipulate heavy objects. Thus the actuation of the stiffness switching was thought to be synchronous with the hand opening/closing DoF and controlled by means the same control channel used by the amputee to control the myoelectric hand.

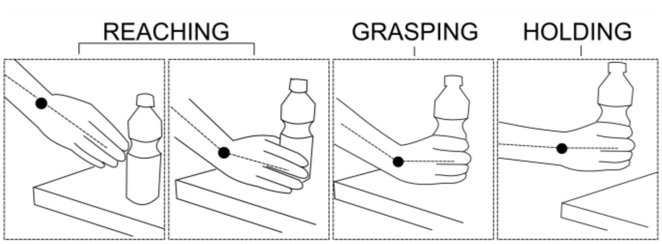

Fig. 2 Typical manipulation sequence.

CONCLUSION

This paper presents the concept of a novel of passive wrist with automatically selectable stiffness. The developed prototype (Fig. 3) weighs only 80 g and its dimensions (diameter = 38 mm, length = 42mm) making the wrist suitable for transradial prostheses at every level of amputation.

Fig. 3 The wrist prototype developed.

ACKNOWLEDGEMENTS

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ADDRESSING THE REIMBURSEMENT CHALLENGE: A SHIFT FROM ADLS TO QOL

Chris Baschuk and Pat Prigge

Upper Limb Prosthetics Society of the AAOP

ABSTRACT

Contemporary upper-limb prosthetic technologies become clinically irrelevant if payers are not willing to reimburse for them. Risk-averse prosthetists are hesitant to embrace and apply newer technologies even when they are the most appropriate choice to provide their patients with the desired functional outcome. These insurance-driven clinical decisions may be one factor in the historically high level of patient rejection of, and dissatisfaction with, upper-limb prostheses.

Exclusionary language is written in many insurance policies regarding upper-limb prosthetic components. A common reason for non-coverage of specific items is that the technology is considered “experimental and investigational” due to a lack of clinical research proving their effectiveness even when they may have been used clinically with success for many years. Multi-articulated hands, powered digit systems, and any prosthesis for an amputation distal to the wrist are most frequently excluded.

As a profession, the focus has been on defining clinical success as meeting ADL requirements. The definition of ADLs used by insurance companies is based on the theoretical independence of a young child. Particularly, it was intended to assess the care needs of elderly persons: including SNF admittance. This is outdated and does not represent upper-limb prosthetic patients’ demands of a pre-injury QOL. There is insufficient clinical evidence specifically quantifying the functional and psychological benefits of contemporary upper-limb prosthetic technologies with respect to improved QOL. Other healthcare fields report and quantify QOL because it provides a broader spectrum in which clinical success is defined. A paradigm shift from assessing and reporting ADLs to QOL in upper-limb prosthetic rehabilitation would help improve our clinical justifications for reimbursement.

The leadership of the Upper-Limb Prosthetics Society of the AAOP is addressing this issue by helping to coordinate and publish research surrounding these contemporary clinical technologies. We have begun to investigate the policies of these insurance companies and tried to determine the requirements that these companies have in place in order for policy guidelines to be changed. The purpose of this presentation is to create awareness surrounding what these requirements are and to initiate a discussion amongst the professionals in attendance at MEC. This is an effort that will need a coordinated international collaboration between manufacturers, clinicians, researchers, physicians, and patient advocacy groups to be successful. Our goal is to establish a body of evidence that can be freely shared amongst those caring for individuals with upper-limb differences so that these prejudicial policies can be overturned.

OUTCOMES OF THE CLINICAL APPLICATION OF PATTERN RECOGNITION IN UPPER LIMB PROSTHETICS: A TWO-YEAR RETROSPECTIVE

Chris Baschuk, Laura Katzenberger, Debra Latour, Thomas Passero and Erik Tompkins

Handspring

ABSTRACT

Presented here is a series of case studies describing the successes and challenges that were experienced, as well as the innovative solutions that were developed, during the real-world clinical application of pattern recognition (PR) technology over the course of a two-year period.

Over the course of two years a total of 13 patients were fit by Handspring Prosthetic Rehabilitation Services with PR technology. Three females and ten males in total. Five patients had a transradial amputation level, seven patients had a transhumeral level amputation, and one patient had a shoulder disarticulation level amputation. One of the patients with a transhumeral level amputation also uses a body powered transradial prosthesis on his contralateral side. One of the patients with a transradial presentation had a congenital limb difference.

Two of the four patients in the transradial group discontinued use of PR. One discontinued use due to general non-compliance, the other discontinued use due to the extra bulk in the prosthesis created by the additional COAPT components.

All of the patients with transhumeral level amputations continue to utilize their PR systems with the exception of the patient with bilateral amputations. This patient was a long-time user of body-powered technology and decided to abandon any attempts at using external powered prostheses.

The one patient with the shoulder disarticulation was initially successful with utilization of the PR technology, but due to health complications secondary to a brachial plexus injury necessitated that the external powered prosthesis be abandoned in favour of a lighter weight custom silicone restoration.

Initially all patients were able to consistently control their prostheses with increased accuracy over the course of their post-delivery occupational therapy.

All patients initially subjectively reported being satisfied with the fit, function, and comfort of their prostheses.

All patients actively utilize the calibration feature of the COAPT system daily when they don the prosthesis for optimal control. Everyone reported that this feature was very important to them.

These case studies demonstrate that the PR technology available from COAPT can be utilized successfully in externally powered prostheses for patients with all levels of upper limb differences. It was the experience of the patients and clinicians at Handspring that the clinical application of PR technology resulted in a 70% myoelectric prosthetic acceptance rate. It was our anecdotal experience that patients fit with the COAPT system were able to progress faster in their OT training than other patients.

PROVIDING HIGH-RESOLUTION TACTILE AND PROPRIOCEPTIVE SOMATOSENSORY FEEDBACK IN HUMANS AFTER LONG-TERM AMPUTATION OF THE HAND

Jacob A. George, David T. Kluger, David M. Page, Suzanne M. Wendelken, Tyler S. Davis, Christopher Duncan, Douglas T. Hutchinson and Gregory A. Clark

The University of Utah

ABSTRACT

The long-term goal of these studies is to provide rich, biofidelic tactile and proprioceptive feedback from an advanced prosthetic hand after prior amputation in humans. Six human subjects (S1-S6) received one or two 100-electrode Utah Slanted Electrode Arrays (USEAs; Blackrock Microsystems) implanted chronically (1-9 months) in residual median and/or ulnar nerves for stimulating sensory fibers (and recording from motor fibers) after long-term (2- to 25-y) transradial amputations. Sensory percepts were mapped by passing increasing current through individual USEA electrodes (biphasic, 200-μs pulses; 100-200 Hz, 200-500 ms trains) until the subject reported a percept (location, type, and intensity), or until stimulation maximum (< 100 μA). Experiments were conducted either in a MuJoCo (Roboti, LLC) virtual reality environment (VRE); or with a simple sensorized, motorized physical prosthetic hand (Open Bionics); or with a more advanced, motorized and sensorized prosthetic hand (DEKA) having 6 DOFs and 19 receptive fields. Subjects reported up to 131 different USEA-evoked cutaneous (e.g., pressure, vibration) or proprioceptive percepts (e.g., joint movement, muscle force). Typically, the evoked percepts covered most of the phantom hand, corresponded to normal afferent fiber distributions, and were enjoyed by subjects. Most percepts showed within-session stability, and in S6 more than half maintained location stability when retested at > 1 month. Subjects successfully discriminated among percepts having different phantom spatial locations or qualities, evoked by individual electrodes or combinations of electrodes. Recent subjects also used sensory feedback evoked by biofidelic afferent fiber stimulation to guide motor control in the VRE. Reciprocally, active engagement with the VRE influenced subjects' perceptions. S6 could discriminate between "soft" foam blocks and "hard" plastic blocks in a sensorimotor task using the DEKA hand (15 successes in 18 trials). S6 also showed objective evidence of embodiment of the simple sensorized, motorized prosthetic hand (as measured by proprioceptive shift from the amputated hand to the prosthetic hand and by responses to survey questions). Stimulation of sensory fibers also resulted in a 23.2% reduction in subjective phantom pain scores for S6 (from 3.75 ± .14 to 2.88 ± .18, $p < 0.005$). Such effects may enhance adoption of advanced hand prostheses by end-users. These results document an unprecedented level of high-resolution tactile and proprioceptive somatosensory percepts in humans with prior hand amputation. The emerging ability to provide a relatively complex repertoire of somatosensory inputs may enhance sensorimotor control, a sense of embodiment, and phantom pain reduction for users of advanced neuroprosthetic limbs.

HOME USE OF A SENSORY RESTORATION SYSTEM: SENSATION STABILITY AND IMPACT ON USAGE

[1]Emily Graczyk, [2]Linda Resnik, [3]Melissa Schmitt and [1]Dustin Tyler

[1]Case Western Reserve University
[2]Providence VA Medical Center
[3]Cleveland VA Medical Center

INTRODUCTION

While neural prostheses to restore sensory feedback to upper limb amputees have the potential to improve task performance and quality of life, studies of sensory restoration systems (SRSs) have only been conducted in controlled laboratory environments. In this study, for the first time, two subjects used a SRS autonomously in a home setting. We report on the technical implementation of the SRS, sensation stability, and participants' attitudes towards and usage of the sensory-enabled prosthesis.

METHODS

Two persons with unilateral trans-radial amputation participated. S1 was implanted with 8-channel Flat Interface Nerve Electrodes (FINEs) around his median and ulnar nerves in May 2012, and S2 was implanted with FINEs around his median and radial nerves in January 2013. The SRS consisted of an Ottobock VariPlus Speed prosthetic hand customized with an embedded aperture sensor and fingertip pressure sensors on D1-D3, an external nerve stimulator with a custom sensory stimulation program, and cabling to connect the stimulator to percutaneous leads. The stimulator mapped pressure signals from the finger sensors into stimulation pulse trains and delivered the stimulation to four electrode contacts on the median nerve.

The five-week ABA crossover study involved two 14-day stages without sensory stimulation (A) surrounding one 7-day stage with sensory stimulation (B). Each day subjects completed surveys on sensory stimulation percepts and reported on their performance of items from a list of everyday activities. On-board usage logs monitored wear time and sensor readings. Interviews were conducted to capture subject perspectives on the SRS. Data was compared across stages to evaluate the effect of sensory feedback.

RESULTS

Subjects were able to independently don and doff the SRS, change stimulation settings, and calibrate the prosthetic sensors. Stimulation parameters and sensation locations remained stable throughout the duration of the study. In stage B, with sensory stimulation, subjects wore the SRS longer (sensation on: 8.4 +/- 3.8 hrs (S1), 8.3 +/- 2.2 hrs (S2); sensation off: 4.7 +/- 2.6 hrs (S1), 6.3 +/- 2.0 hrs (S2)), used it more frequently to touch/manipulate objects (S2: $p=0.03$), and reported using their prosthesis to do more activities (S1: $p=0.03$; S2: $p<0.001$). Participants preferred using the prosthesis with sensation enabled.

CONCLUSIONS

Two trials of a take home SRS were successfully completed and demonstrate initial feasibility. The SRS was well-received. Interviews and usage logs indicated that subjects preferred using the prosthesis with sensory feedback. Robustness, reliability, and ease of use are critical design features for an SRS.

HIGH-RESOLUTION, REAL-TIME MOTOR CONTROL OF PROSTHETIC HANDS IN HUMANS AFTER LONG-TERM AMPUTATION

Gregory Clark, Suzanne Wendelken, Tyler Davis, David Kluger, David Page, Jacob George, Christopher Duncan and Douglas Hutchinson

University of Utah

ABSTRACT

To explore the ability to use peripheral neural and myoelectric signals to control advanced prosthetic hands, six human subjects (S1-S6) received one or two 100-electrode Utah Slanted Electrode Arrays (USEAs; Blackrock Microsystems) implanted chronically (1-9 months) in residual median and/or ulnar nerves for recording from motor fibers and for stimulation of sensory fibers (George et al., MEC17) after long-term (2- to 25-y) transradial amputations. S5 and S6 also received a 32-electrode electromyogram (EMG) assembly implanted in residual forearm muscles (Ripple, LLC). Motor control was provided by real-time decodes of myoelectric and neural signals; myoelectric signals provided the dominant control in subjects with both implants. EMG power and neural firing rate provided the features used for Kalman-filter decode algorithms. During initial "training" sessions, subjects viewed individuated digit or wrist movements of a virtual hand and attempted to mimic these movements with their phantom hand. The neural and EMG activity associated with the imagined phantom movements was then used to select neural and EMG channels from among 720 single-ended or differential possibilities (Nieveen et al., MEC17), and to set the parameters of the Kalman filter. The Kalman filter output was used to control a virtual or physical prosthetic hand in subsequent "testing" sessions. Experiments were conducted either in a MuJoCo (Roboti, LLC) virtual reality environment; or with a simple sensorized, motorized physical prosthetic hand (Open Bionics); or with a more advanced, motorized and sensorized prosthetic hand (DEKA) having 6 DOFs and 19 receptive fields. Recent subjects successfully controlled up to nine real-time degrees-of-freedom (DOFs) involving 18 digit and wrist movements of a virtual hand in a formal target-touching test (e.g., 53 successes in 54 trials). One subject achieved up to 12 apparent DOFs in informal tests. Both proportional position and velocity control were achieved. Additionally, subjects successfully combined individual DOF movements into novel grasps (e.g., "pinch") that had not been explicitly trained. EMG decodes remained stable for over a week (e.g., 26 successes in 26 trials in a 3-DOF, 4-level novel virtual grasp-matching task). S6 controlled the digits and wrist of the DEKA physical hand in both trained and untrained movements and grasps. Subjects also successfully used USEA-evoked sensory feedback to guide their motor behaviors in real-time closed-loop control. These results document an unprecedented level of real-time proportional control of a prosthetic hand in humans with long-term hand amputation. Future research includes translating these approaches to a wireless, practical take-home system.

Layperson's 3-D Printed Post-Operative Prostheses Following Bilateral Wrist Disarticulation

Keaton Valentine, Trevor Valentine, Phil Stevens MEd, CPO

Hanger Clinic

ABSTRACT

This paper presents novel 3-D printed, post-operative prostheses created by the brother of a patient who had undergone bilateral wrist disarticulation amputations to span the time period between amputation and the fitting of preparatory prostheses. Following severe frost bite, attempted limb salvage and eventual amputation, the patient had been without upper limb prehensile function for 5 weeks at the time of his initial prosthetic consult. Following removal of surgical sutures, as the limbs continued to heal and volume reduction efforts were implemented, the patient's brother devised and manufactured post-operative prostheses to restore a degree of prehensile function over the next several weeks until the patient was fitted with preparatory body powered devices. The combination of 3-D printed and commercially available elements enabled the patient to hold and preposition utensils and paper work. In a separate configuration, he could hold a smart phone with one limb while using a stylus attached to the contralateral limb to navigate the phone screen. Elements of these designs will be described. The role of 3-D printing in the addressing the light duty, short term, immediate needs of post-operative prostheses may warrant further consideration and development.

INTRODUCTION

The "golden period," described by Malone et al [1] as the first 30 days following amputation, has been suggested as the ideal time window to introduce upper limb prostheses. For patients with unilateral limb loss, this window is thought to influence prosthetic acceptance and compliance as the longer the time between amputation and prothetic rehabilitation, the more skilled the individual may become at functioning as a one-handed individual. For bilateral patients, the value of prosthetic fitting during this golden period is perhaps more direct, related simply to the restoration of upper limb function and some level of independence. This cases study presents a case of bilateral wrist disarticulation in which a family member devised and manufactured simple post-operative prostheses to restore limited independence in the days following suture removal.

INITIAL PRESENTATION

The individual presented in this case, TV, experienced extreme bilateral frostbite of the hands bilaterally when he was caught outdoors overnight in an unexpected snowstorm. Nearly 4 weeks intervened between the initial frost bite and the wrist disarticulation amputation. The patient's initial prosthetic consultation occurred 5 weeks after the initial injury and 10 days after the amputation (Figure 1). At this time, the patient reported that he was able to don and doff a T-shirt with difficulty, but was unable to wash his face, comb his hair, put on socks, tie shoes, bathe, prepare a light meal, drink from a cup, toilet independently, use kitchen utensils, write or use a smart phone.

The patient was advised to begin compression therapy following suture removal. A treatment plan for bilateral preparatory prostheses was developed.

Figure 1: Residual limbs post-amputation, prior to suture removal

POST-OPERATIVE PROSTHESES

The patient returned for casting, fabrication and delivery of bilateral, preparatory body-powered prostheses 3 weeks later, following suture removal and two weeks of compressive garment therapy. At this appointment, he presented with a modular post-operative prosthesis on his right limb, devised and fabricated by his brother using both 3D printed and commercially available elements (Figure 2).

Figure 2: Right post-operative prosthesis in a cell-phone mount and prehensile device configuration

The right device consisted of a custom gutter-style socket with an integrated wrist base and a distal 1.5” ball adaptor. This element was created by the patient’s brother using a series of programs including Sketchup, Autocad and Makerbot print (Figure 3). Anatomic length and width measurements were used to configure the dimensions of the gutter splint.

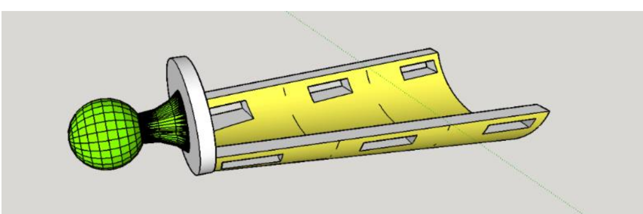

Figure 3: CAD rendering of the right gutter-style splint

Following printing out of PLA on a Maker Bot Replicator +, this element was lined with a compressive foam and simple strapping was configured (Figure 4).

Figure 4: Printed version of the gutter splint with straps

The prehensile device was also created by the patient’s brother using a series of programs including Sketchup, Autocad and Makerbot print. The device consisted of opposing tines that articulate through a hinge at their base. The base of one tine is integrated with a 1.5” ball. The base of the other tine is integrated to an exaggerated thumb lever (Figure 5).

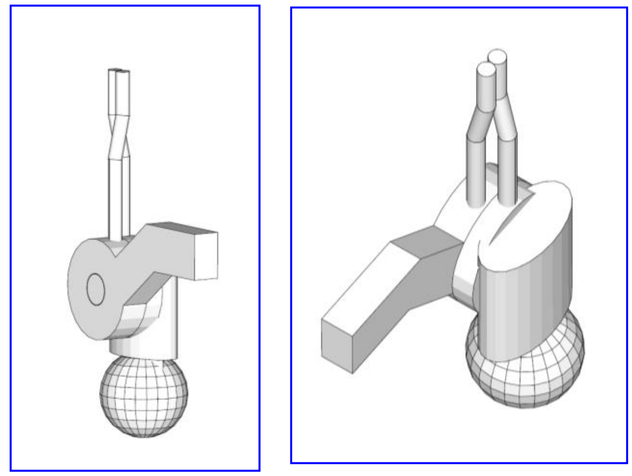

Figure 5: CAD rendering of the 3D printed prehensile device

Once printed from PLA on a MakerBot Replicator + printer, household rubber bands were used to create the voluntary opening prehensile force. The exagerated thumb of the hook allowed the patient to open the hook with his contralateral limb (Figure 6).

Figure 6: Printed prehensile device with rubber bands providing closing force

These two elements were joined through a commercially available RAM Mount. This element accepts the 1.5” balls of both the gutter splint and the hook with a variable tension compression system that regulates the friction within the twin ball-in-socket joints (Figure 7)

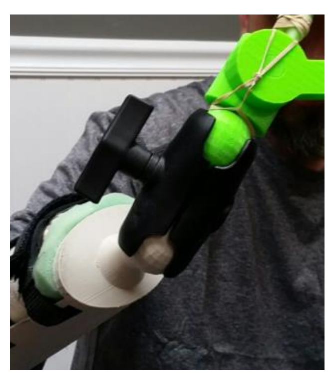

Figure 7: RAM Mount connecting adapter

In a separate configuration, the 1.5" ball of a commercially available smart phone mounting frame can be installed in the distal socket of the RAM Mount. This allows the patient to orient the phone for reading. A stylus attached to the left limb with a simple strapping system permits the patient to navigate activities on the smart phone (Figure 8)

Figure 8: Smart phone configuration with commercially available RAM Mount attached to the right prosthesis and a stylus attached to the left limb

This device effectively spanned the final 3 weeks of the one month gap between amputation and receipt of his preparatory prostheses, restoring a small measure of functional independence.

PREPARATORY PROSTHESES

One month after his amputation, the patient was fit with a more definitive solution in the form of preparatory body powered prostheses (Figure 9).

Figure 9: Preparatory body-powered prostheses

OBSERVATIONS AND RECOMMENDATIONS

When asked to contribute to this abstract, the designer of this post-operative prosthesis volunteered the following observations:

"The transition period between amputation and the fitting of "permanent" prosthetics is … a crucial time for an amputee to physically and mentally cope with the loss of a limb or limbs….there were no intermediate devices that I was aware of that solve the problems of a person in his situation. In the hospital, we had a roll of tape and a stylus to work with. It was very hokey looking, and a sad snapshot of the current state of awareness for amputees. I know that simple 3D printed devices most likely are not a long term solution, but there is value in a short term answer for the new issues that have developed. I hope that in the near future there will be options available in hospitals for amputees that can address their immediate needs and wants."

While the strength and durability of entry-level PLA-printed devices are likely ill-suited for long term use or typical activities of daily living, they may represent a reasonable medium for the fabrication of temporary, light duty devices used to restore a measure of functional independence in the days following upper limb amputation.

As this case study was reported retrospectively, no formal informed consent documentation was prospectively obtained. However, a written media release was signed by the patient prior to the submission of this abstract.

REFERENCES

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MIRROR THERAPY IS EFFECTIVE FOR TREATING UPPER EXTREMITY PHANTOM PAIN

[1]Sacha Finn, [2]Briana Perry, [3]Jay Clasing, [3,4]Lisa Walters, [3]Sandra Jarzombek, [5]Sean Curran, [6]Minoo Rouhanian, [7]Paul Pasquina and [8]Jack Tsao

[1]UCSF
[2]UCSD
[3]BAMC
[4]Advanced Arm Dynamics
[5]CNRM
[6]WRNMMC
[7]USUHS
[8]University of Tennessee Health Science Center

OBJECTIVE

Phantom limb pain is prevalent in patients post-amputation and is difficult to treat. We assessed the efficacy of mirror therapy in relieving phantom limb pain in unilateral, upper extremity (UE) amputees.

METHODS

Fifteen participants from Walter Reed and Brooke Army Medical Centers were randomly assigned to one of two groups: mirror therapy (n=9) or control (n=6, covered mirror or mental visualization therapy). Participants were asked to perform 15 minutes of their assigned therapy daily for four weeks. The primary outcome was pain as measured using a 100-mm Visual Analogue Scale (VAS).

RESULTS

Subjects in the mirror therapy group had a significant decrease in pain scores, from a mean of 44.1 (SD=17.0) to 27.5 (SD=17.2) mm (p=0.002). In addition, there was a significant decrease in daily time experiencing pain, from a mean of 1022 (SD=673) to 448 (SD=565) minutes (p=0.003). In contrast, the control group had neither diminished pain (p=0.65) nor decreased overall time experiencing pain (p=0.49). A response seen by the tenth treatment session was predictive of final efficacy.

CONCLUSIONS

These results confirm that mirror therapy is an effective therapy for phantom limb pain in unilateral, upper extremity amputees, reducing both severity and duration of daily episodes.

INTRODUCING A NOVEL TRAINING AND ASSESSMENT PROTOCOL FOR PATTERN MATCHING IN MYOCONTROL: CASE-STUDY OF A TRANS-RADIAL AMPUTEE

Markus Nowak[1], Raoul M. Bongers[2], Corry K. van der Sluis[2] and Claudio Castellini[1]
[1]*DLR – German Aerospace Center, Weßling, Germany*
[2] *University Medical Center, University of Groningen, The Netherlands*

BACKGROUND AND AIM

Multi-DoF prostheses and advanced myocontrol challenge both engineers and rehabilitation professionals to teach the patient to optimally control the prosthesis, and to assess their addition to functional recovery. This work proposes and evaluates a standardized clinical procedure of

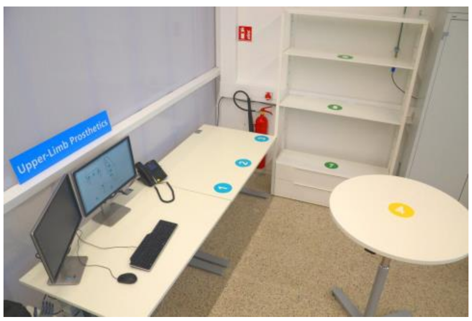

Figure 1: the laboratory environment.

a training and assessment protocol characterized by reciprocal adaptation of subject and prosthesis in a case-study.

MATERIALS AND METHODS

One 35yrs old male trans-radial left-hand amputee (amputation in 2005), routinely using a *Variplus* hand (Otto Bock GmbH) with standard two-electrode control since 2012. For the experiment, he was fitted with a customized socket (Pohlig GmbH) with eight 13E200 *MyoBock* sensors (Otto Bock GmbH) and an *i-LIMB Revolution* prosthetic hand (Touch Bionics, Ltd.).

The protocol was organised in *sessions*, each session spanning several visits, characterised by patterns *(actions)* to be recognised by the myocontrol system and by *tasks* to be performed.

session 1: rest, power grasp
session 2: rest, power grasp, precision grip
session 2': rest, power grasp, pointing index
session 3: rest, power grasp, pointing index, thumbs up

The required tasks varied from performing simple grips to daily living tasks, using associated actions in a room-sized laboratory (see **Figure 1**). Non-linear incremental regression enforced interactive simultaneous / proportional control, so that *on-demand updating* could be applied. Measures of performance were time required to complete each task, and the number of required updates.

Along six months, we measured four sessions of increasing difficulty, determined by using body postures, distances to be walked, graded-force tasks, etc. The subject's satisfaction was assessed multiple times per session, and the tasks were adjusted accordingly.

RESULTS

Fisher's index, measuring data cluster separation, applied to the subject's signals per each action, is visible in **Figure 2**, where each data point represents a visit. The colour indicates the session. Sessions 1 and 2' showed a stable performance after several visits (circles), while sessions 2 and 3 (square) did not reach a reliable. Fisher's

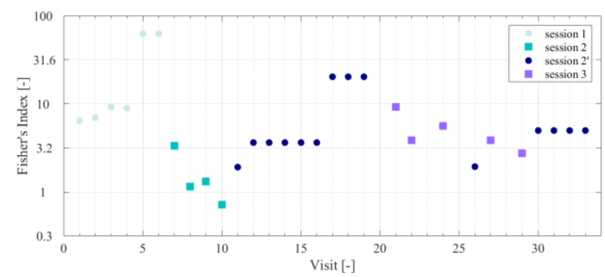

Figure 2: Fisher's index of the subject's signals over time.

index increased during sessions 1 (visit 1-6) and 2' (visit 11-19, 26, 30-33). The subject's satisfaction increased over sessions.

DISCUSSION

Measurements obtained during the experiments can *and should* also be exploited as a guidance for the experimenter: for instance, in session 2 we introduced a new action which proved to be unfeasible for the subject; the no-increase in Fisher's index induced us to switch to another action, which produced a definite improvement in session 2'. The findings indicate the feasibility of the protocol.

ACKNOWLEDGEMENT

This work was partially supported by the German Research Agency project *TACT-HAND: Improving control of prosthetic hands using tactile sensors and realistic machine learning* (DFG Sachbeihilfe CA1389/1-1).

REAL-TIME PROPORTIONAL MYOELECTRIC CONTROL OF DIGITS

Agamemnon Krasoulis[1,2], Sethu Vijayakumar[1], Kianoush Nazarpour[3, 4]

[1]*Institute of Perception, Action and Behaviour, School of Informatics, University of Edinburgh, UK*

[2]*Institute for Adaptive and Neural Computation, School of Informatics, University of Edinburgh, UK*

[3]*School of Electrical and Electronic Engineering, Newcastle University, UK*

[4]*Institute of Neuroscience, Newcastle University, UK*

ABSTRACT

Current powered hand prostheses offer the potential of individual digit control. In practice, however, sequential control strategies are employed (e.g. grip control) which lead to under-actuation of the prosthesis. Arguably, adoption of proportional control strategies can improve the ease of use and naturalness of upper-limb prostheses. Previous work has demonstrated the feasibility of proportional finger control with invasive methods, that is, by using intra-muscular electrodes. Although it has been shown that it is feasible to reconstruct finger joint angle trajectories offline by using surface electromyography (sEMG), the real-time efficiency of such control systems has not been previously evaluated. In this study, we implemented a real-time, proportional finger joint angle controller for the 5 degree-of-freedom (DOF) IH2 Azzurra anthropomorphic hand, and tested its performance with ten able-bodied and two trans-radial amputee subjects. Myoelectric activity was recorded on the participants' forearm proximal to the elbow, by using 16 sEMG sensors, which were equally spaced around the subjects' forearm. The recorded EMG activity was used to decode the intended degree of individual finger flexion and thumb opposition by using a linear regression system (Wiener filter), thus mapping muscle activity to 5-DOF finger joint angles. Ground truth data were collected by using a data glove placed on the participants' contralateral hand. Two different sets of experiments were performed for both populations of participants. In the first experiment, participants were asked to modulate their muscular activity in order to control the robotic hand such that its posture matched a target posture presented to them on a computer screen. At the end of each trial, participants received a performance score that was based on the L_1-distance between the target and performed postures. The analysis of results from this experiment provides helpful insights into the mechanisms underlying the learning of controlling a 5-DOF robotic hand. In the second experiment, participants were asked to control the artificial hand to perform a pick-and-place task, again by modulating their muscular activity. Preliminary results from this part of the study shed light on the usability of proportional finger control for performing activities of daily living (ADL). We value the findings of our study as a valuable step towards the development of truly dexterous, non-invasive hand prostheses.

REAL-TIME CLASSIFICATION OF FIVE GRIP PATTERNS WITH ONLY TWO SENSORS

Agamemnon Krasoulis[1,2], Kianoush Nazarpour[3, 4], Sethu Vijayakumar[1]

[1]*Institute of Perception, Action and Behaviour, School of Informatics, University of Edinburgh, UK*

[2]*Institute for Adaptive and Neural Computation, School of Informatics, University of Edinburgh, UK*

[3]*School of Electrical and Electronic Engineering, Newcastle University, UK*

[4]*Institute of Neuroscience, Newcastle University, UK*

ABSTRACT

Current state-of-the-art prosthetic hand systems utilize electromyography (EMG) signals recorded from the user's residual muscles to decipher movement intention and control the prosthesis. To achieve a high level of decoding accuracy and robustness, a large number of EMG sensors is typically required. This requirement can both limit the functionality of the device from the user's perspective, as well as increase the power/computational requirements of the system. In this study, we propose a full framework for efficient and robust pattern recognition-based prosthetic hand control by using a single pair of EMG/inertial measurement (IM) sensors. Our proposed framework can be summarised as follows: 1) identify the optimal sensor location during an initial screening session by using a standard sequential forward selection algorithm; 2) deploy a regularized version of discriminant analysis classification which we have found that greatly outperforms linear discriminant analysis (LDA) when the input feature dimensionality is small; 3) adopt a novel classification rejection algorithm to minimize the controller's false positive rate (FPR). We assessed the performance of our proposed framework by conducting a real-time pick-and-place experiment with twelve able-bodied and two trans-radial amputee subjects Five different hand grips were included in our experiments: power/cylindrical, lateral/key, tripod, index pointer, and hand open. We found that after a few trials, participants were able to achieve robust prosthetic control performance (95% and 85% completion rates for able-bodied and amputee subjects, respectively). Furthermore, completion times were comparable to our previous work, where a larger number of sensors were used (4-6). This study provides a proof-of-principle for efficient pattern recognition-based prosthetic hand control with existing two-site EMG clinical systems.

PROPORTIONAL AND SIMULTANEOUS ESTIMATION OF COMBINED FINGER MOVEMENTS FROM HIGH-DENSITY SURFACE EMG

[1]Sigrid Dupan, [2]Michele Barsotti, [3]Ivan Vujaklija, [4]Dick Stegeman, [5]Bernhard Graimann, [3]Dario Farina and [6]Strahinja Dosen

[1]Department of Biophysics, Donders Institute for Brain, Cognition, and Behaviour, Radboud University, Nijmegen
[2]PERCRO lab, Scuola Superiore Sant'Anna, Ghezzano
[3]Department of Bioengineering, Centre for Neurotechnology, Imperial College London, London
[4]Department of Neurology/Clinical Neurophysiology, Donders Institute for Brain, Cognition, and Behaviour, Radboud University Nijmegen Medical Centre, Nijmegen
[5]Department of Translational Research and Knowledge Management, Otto Bock HealthCare GmbH, Duderstadt
[6]Neurorehabilitation Systems Research Group, Clinics for Trauma, Orthopaedics and Plastic Surgery, University Medical Center Goettingen, Goettingen

BACKGROUND

In order to mimic the dexterity of the human hand, modern hand prostheses allow control of individual fingers. However, proportional and simultaneous control of individual fingers using myoelectric signals is still to be achieved. For practical application, it is important to minimize the number of sensors and the amount of calibration data. The aim of the present study was to test the feasibility of estimating finger forces during both single and combined finger movements using a training set of only 6 motions and a reduced set of electrodes.

METHODS

Five subjects performed 19 flexion movements: 5 individuated finger flexions, and 14 combinations of 2 to 5 fingers. The finger forces were measured using the Amadeo robot (Tyromotion GmbH, AT), and the surface electromyography (EMG) was recorded through a 256-channel high-density electrode grid (OTBioelettronica). Ridge regression was applied to simultaneously predict the forces of all fingers. The only movements used in the training were the single finger flexions, and the simultaneous flexion of all fingers. The quality of estimation was evaluated by computing the root mean square error (RMSE) between the estimated and desired force normalized (nRMSE) to the maximum of the desired force. The regression was performed using a full set of electrodes and reduced sets comprising 48 and 24 electrodes. The statistically significant difference was tested using a one-way ANOVA ($p<0.05$).

RESULTS

The average nRMSE for the training data and full electrode set was 0.12±0.03, and there was no significant difference in the quality of estimation between the fingers. However, the performance decreased ($p<0.001$) when using less electrodes (48 electrodes: 0.20±0.04; 24 electrodes: 0.22±0.04). The estimation of the finger forces during combined motions (test data) resulted in nRMSE of 0.40±0.12, and the performance was significantly better ($p=0.003$) for the ring finger (0.33±0.08) compared to the thumb (0.49±0.12). There was no difference in the quality of estimation for the combined motions when reducing the number of electrodes.

CONCLUSION

The study showed that the finger forces can be estimated proportionally and simultaneously using a simple method and a reduced training set. However, the average precision of estimating combined finger movements was not high, and the estimation of the thumb proved to be the most difficult of all the fingers. This likely reflects an increased role of intrinsic muscles in thumb control. Remarkably, reducing the number of electrodes did not significantly decrease the performance for the combined movements.

A PRELIMINARY STUDY TOWARDS AUTOMATIC DETECTION OF FAILURES IN MYOCONTROL

Markus Nowak, Sarah Engel and Claudio Castellini
DLR – The German Aerospace Center, Weßling, Germany

ABSTRACT

Reliability is still the main issue in myocontrol: enforcing (dexterous) grasping, releasing and moving exactly and only when the wearer desires it. One specific path towards the solution of this problem is incremental machine learning, leading to *interactive myocontrol*, in which unreliability is taken care of via on-demand model updates, requested by the experimenter and/or the subject herself/himself. One natural drawback of this approach is that an "oracle" is needed at all times, stopping the prediction and calling for an update whenever this is deemed to be the case; an automated oracle, as reliable as possible, is therefore very desirable.

This work shows the results of a preliminary study in which we tried to find features of the control signals and predictions, as well as environmental information (inertial sensors and motor currents) to automatically identify the failures of the myocontrol system. The outcome is promising, showing that a classifier can match the observer's judgement with an overall average accuracy of slightly more than 75%.

INTRODUCTION

Whenever the scientific community talks about dexterous myocontrol, i.e., natural, simultaneous and proportional (s/p) control of prosthetic artefacts over many degrees of freedom (DoFs), the main issue remains that of *reliability*. An unreliable myocontrol system lets the prosthesis open, close and grasp at times when such an action is not required and vice-versa, which can lead to disastrous results. Even in the case of traditional two-sensors EMG-based myocontrol the situation is far from optimal, mainly due to unexpected changes in the signals (caused by sweating, displacement, fatigue, etc.). There is still a lot of work to do, as has recently been shown during the Cybathlon ARM competition[1]: the winner of the competition, Robert Radocy of TRS Prosthetics, was using a body-powered one-DoF prosthetic arm, which enabled him perform all required tasks without any error, swiftly and elegantly; and he was competing against some of the most advanced academic solutions in the world.

Still, there is now plenty of surveys [Micera et al. (2010), Peerdeman et al. (2011), Ison and Artemiadis (2014), Engdahl et al. (2015)] showing that advanced control is desired, but it is rejected due to poor reliability. Our way towards the solution of the problem is *incremental learning*, allowing for on-demand model updates in real time, leading to *interactive myocontrol:* a natural, s/p control schema which can be "taught" new information whenever the experimenter and/or the subject deem it necessary [Gijsberts et al. (2014), Strazzulla et al. (2016), Nowak and Castellini (2016)]. This concrete possibility of updating represents the main strength of interactive myocontrol; however, the necessity of having an "oracle" at one's disposal – be it the experimenter or the subject – probably constitutes its main weakness.

In this work we propose a step towards automatic updating of interactive myocontrol. In particular, we show that specific features extracted from either surface electromyography (sEMG) signals, the predicted control commands, inertial and/or current measurements, can be used to characterise when dexterous myocontrol would fail. In a preliminary analysis, we engaged an intact subject, equipped with a commercial sEMG bracelet and a multi-fingered 6-DoFs prosthesis, in a complex series of daily-living tasks inspired by the Cybathlon ARM race concourse; the reliability of the myocontrol system would be stressed through the usage of diverse actions (grasping patterns) as well as the necessity to move and walk around. In this specific experiment, an offline analysis reveals that a standard linear classifier could discriminate the faulty situation in 76.71% of the cases.

MATERIALS AND METHODS

Experimental Setup

The experimental setup is visible in Figure 1. It consists of (a) a commercial orthotic splint that was fitted with a custom-design mounting for prosthetic hands; (b) an *i-LIMB Revolution* multi-fingered prosthetic hand manufactured by Touch Bionics; (c) a *Myo* bracelet by Thalmic Labs, embedded with eight sEMG sensors covering the full circumference of the user's proximal forearm. The *i-LIMB* also provides the motor current readings and the "digit

[1] see http://www.cybathlon.ethz.ch/en/cybathlon-news/cybathlon-results/arm-results.html and, especially, the video excerpts in http://www.swisswuff.ch/tech/?p=6670.

status" (opening, closing, stalled), while the *Myo* mounts an inertial device providing the translational and angular acceleration in three dimensions.

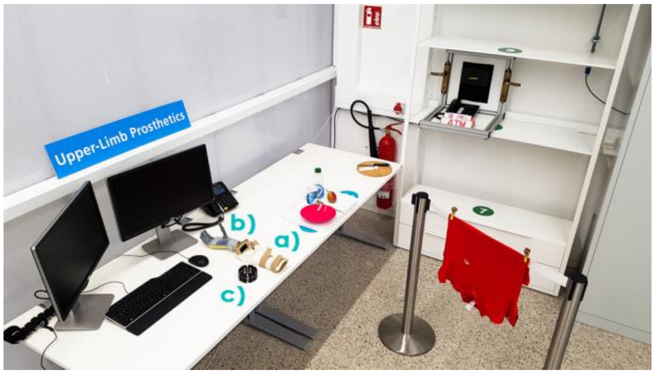

Figure 1: overview of the laboratory setup including a) the *i-LIMB* on an b) orthotic splint, c) the *Myo* bracelet and daily-living objects that were manipulated during the experiment.

Incremental s/p myocontrol was enforced using six parallel instances of *Ridge Regression with Random Fourier Features* (RR-RFF), a method already tested and used for myocontrol in, e.g., [Gijsberts et al. (2014), Strazzulla et al. (2016)]; sEMG data with mild low-pass filtering was taken as the input space, while the output space (ground truth) was obtained through on-off goal-directed visual stimuli administered to the subject (see again the cited references, plus [Sierra González and Castellini (2013)]). The six outputs of the RR-RFF instances were directly fed as (proportionally scaled) current commands to the six motors of the prosthetic hand.

Experimental Protocol

The protocol enforced the execution of a series of nine daily-living tasks inspired by the *SHAP* assessment protocol [Light et al. (2002)] as well as by the Cybathlon ARM competition (see Table 1); notice that the tasks involve walking, standing, sitting and moving around – to this aim, a predefined path was arranged in our laboratory (see Figure 1 again). Figure 2 shows a schematic depiction of our own "concourse", built within the laboratory.

Table 1: tasks involved in the experimental protocol.

Task#	Task description
1	Place objects on a tray
2	Carry a tray with objects on it
3	Place objects on shelves of different heights
4	Cut a mock-up cucumber
5	Pour mock-up water in a mug
6	Hang a piece of clothing on a clothesline using pegs; take it down; fold it
7	Withdraw money at a mock-up ATM
8	Unwrap a piece of candy
9	Shake hands with the experimenter

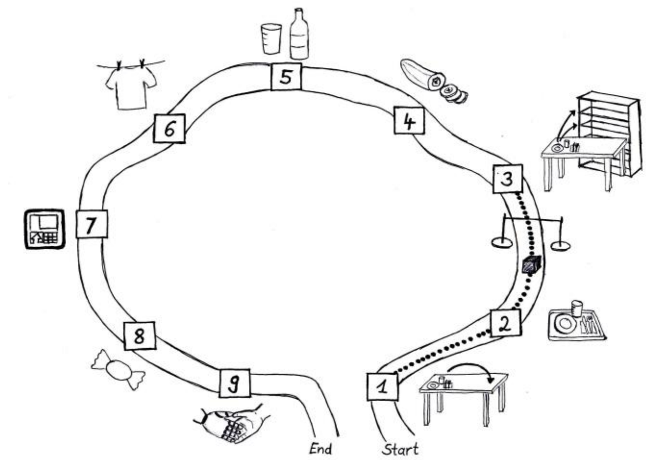

Figure 2: a schematic depiction of the path through the laboratory, including pictograms denoting each task.

The protocol required the subject to use three different hand configurations *(actions)*, namely power grasp, tripodal grasp (precision) and pointing index. Initially, the user was asked to perform only one repetition of all these actions (plus the resting action) to train the myocontrol system; such a low number of repetitions was explicitly chosen to potentially induce instability and poor performance in the control system during the execution of the protocol.

During the execution of each task the user was closely observed by the experimenter; the performance of the control system was marked online as *good* or *poor* by the experimenter. When the performance was considered *good*, e.g. the task was successfully completed, the user moved on to the next task; in case the performance was considered *poor*, e.g., objects were dropped or the prosthesis did not behave the intended way, the user was asked to continue to try for a while; then, additional sEMG data was gathered for the intended action (on-demand model update – each update took approximately 15s). After each update the last action was repeated, then the whole task was carried on until successfully performed. This procedure allowed us to gather both *good* and *poor* performance labels even within one and the same task.

Before the experiment started, the single subject signed an informed consent form. The experiment was approved by the Work Safety Committee of the DLR and it was performed according to the declaration of Helsinki.

Observer Model and Feature Extraction

To try and automatically determine the quality of the performance, that is to mimic the experimenter+subject "observer", we offline fed features extracted from the sEMG signals, the myocontrol predictions, the acceleration, the motor current and the digit status signals to a standard linear classification method (*Linear Discriminant Analysis*, see Figure 3).

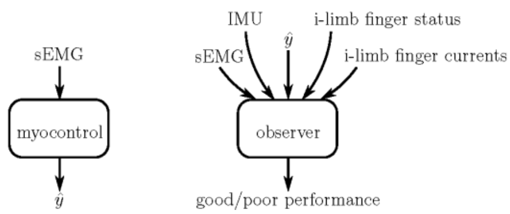

Figure 3: signals fed to the myocontrol system and observer model. In the Figure, $\hat{y}$ denotes the (six-dimensional) output of the myocontrol system itself.

A reasonable assumption is that the myocontrol is unstable whenever its prediction (in turn depending on the input data) displays an oscillatory behaviour; therefore, a Fast Fourier Transformation (FFT) was applied to the sEMG signals as well as to the predictions; the FFT coefficients were then reduced to one using Principal Component Analysis (PCA). Furthermore, a threshold was placed on the derivative of the inertial signals to determine the status of acceleration. The *i-LIMB* digit status was analysed using a derivative and a subsequent count of the zero-crossings, over a moving window of 0.5s. Lastly, FFT was applied to finger currents and reduced to three dimensions, again using PCA. “Leave-one-task-out” cross validation was applied to the extracted features to train the LDA (the observer was trained on all but one tasks and tested on the remaining tasks). This resulted in a 20-fold cross validation, since the user performed 20 tasks in total, including the repetitions of failed tasks.

RESULTS

The classifier showed an overall average accuracy of 76.71%. Since in this preliminary study only one subject was examined, no statistical or comparative analysis could be performed; still, qualitative inspection (an example is found in Figure 4) reveals that some of the features extracted from the available data uniformly match the *good* / *poor* performance, as well as the prediction of the observer evaluating the performance.

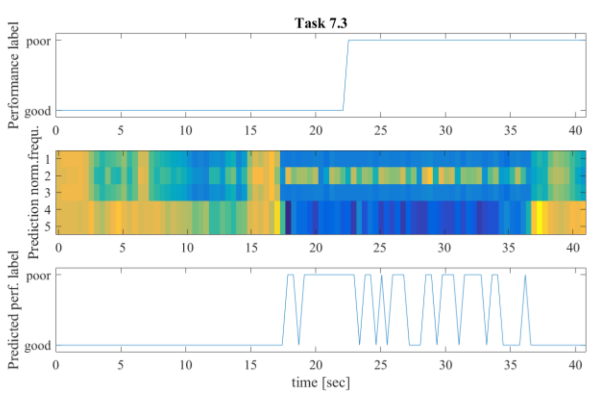

Figure 4: exemplary labels and features for task 7, 3rd trial: labels (top panel); features extracted from $\hat{y}$ (middle panel); labels predicted by the classifier (bottom panel).

Especially, as we expected, features somehow representing the degree of oscillation in the sEMG, prediction and motor current signals seems to match the *poor* performance. This is easily interpreted as the prediction system finding itself in an ambiguous situation (unexpected changes in the input signals, for instance).

More in detail, Table 2 shows the number of attempts performed, and the classifier accuracy, per each task.

Table 2: number of attempts and accuracy in the prediction of the classifier for each task.

Task#	Attempts	Accuracy (mean±std)
1	3	57,15% ± 28,14%
2	2	79,74% ± 13,46%
3	2	65,45% ± 25,59%
4	1	98,15%
5	3	80,67% ± 17,20%
6	1	97,10%
7	6	67,99% ± 16,13%
8	2	60,61% ± 3,85%
9	1	100%

DISCUSSION AND CONCLUSION

Discussion

To a large extent, *poor* performance of the myocontrol, as identified by the experimenter and/or the user, could be automatically identified in more than three quarters of the cases. This was obtained based on sensor information that is already present in advanced myoelectric control, with the exception that most systems lack an inertial sensor.

Arguably, we assume the reported accuracy of the observer is lower than the actual percentage. In Figure 4 one can see the manually labelled performance, the features of the prediction $\hat{y}$ and the labels predicted by the observer. From this figure one can see the delay of the experimenter in labelling the data. While the features and the observer already correctly label the performance as *poor*, there is a reaction time of the experimenter, who draws her/his conclusion based on visual information only.

Conclusion

This results makes us confident that having an observer of the myoelectric performance will provide information on the status of the prosthetic control and therefore allow the user to interact with the control and improve it where needed.

This is a first step towards a truly interactive prosthetic control, where the system can identify shortcomings of itself and ask the wearer for guidance. An example would be a ML-based myocontrol that has been training in a sitting position of the user, who then continues to manipulate objects on a high shelf. Due to changes in the muscular configuration the sEMG signals might be different form the training data and therefore result in a poor prediction. The system could recognise said poor performance and interact with the user to improve the myocontrol.

Postural variations have been identified as a source of poor performance [Fougner et al. (2012)] and to resolve this issues excessive training in different positions coving most of the assumed workspace of a prosthesis was applied to gather as much sEMG data as possible. But one can only train in so many positions. Our work has a similar goal, but the approach is fundamentally different. We only ask for a short initial round of calibration, which is only updated by new data upon demand. This reduces the initial training burden and provides the user with a highly interactive way of controlling her or his prosthesis.

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MULTI-LEVEL COMBINATION OF ELECTROMYOGRAM AND INERTIAL MEASUREMENTS FOR IMPROVED MYOELECTRIC PATTERN RECOGNITION

Rami Khushaba[1], Agamemnon Krasoulis[2,3], Kianoush Nazarpour[4, 5]

[1]*Faculty of Engineering and IT, University of Technology, Sydney, Australia*

[2]*Institute of Perception, Action and Behaviour, School of Informatics, University of Edinburgh, UK*

[3]*Institute for Adaptive and Neural Computation, School of Informatics, University of Edinburgh, UK*

[4]*School of Electrical and Electronic Engineering, Newcastle University, UK*

[5]*Institute of Neuroscience, Newcastle University, UK*

ABSTRACT

Advanced pattern recognition-based myoelectric prosthetic hands are currently limited due to the inadequate real-time control performance and the lack of classification robustness. In one direction of research, the extraction of accurate and efficient descriptors of muscular activity has been a major focus for improved myoelectric control while another direction considers the electromyogram (EMG) signal inadequate for reliable control and suggests that the use of inertial measurement (IM) data is needed. We propose to address the current limitations by considering a combination of robust feature extraction methods and a fusion of EMG and IMs. Our feature extraction algorithm employs the orientation between a set of descriptors of muscular activities and a nonlinearly mapped version of it. It also shifts the voting step from the classifier to the feature extraction stage by fusing the EMG signal power spectrum characteristics derived from each analysis window with the descriptors of previous windows for robust activity recognition. The proposed idea can be summarized in the following three steps: 1) extract power spectrum moments from the current analysis window and its nonlinearly scaled version in time-domain through Fourier transform relations, 2) compute the orientation between the two sets of moments, and 3) apply data fusion on the resulting orientation features for the current and previous time windows and use the result as the final feature set. We collected and analysed a dataset comprising of 20 able-bodied and two amputee participants executing 40 movements. In our experiments, we firstly show that the well-known methods can only achieve an average of 25% classification error across all subjects with 150 ms windows, and by using our proposed features we achieved significant reductions in error rates of up to 16% across all subjects ($p < 0.001$). The inclusion of the IM data further significantly enhanced the results by shrinking the classification error rates to an average of < 5% across all subjects ($p < 0.001$). We consider that the implications of our study could help improve the usability of upper-extremity prostheses in real-life applications.

IMPROVING OPTICAL MYOGRAPHY VIA CONVOLUTIONAL NEURAL NETWORKS

Christian Nissler*, Imran M. Badshah*, Claudio Castellini*, Wadim Kehl² and Nassir Navab²

**Institute of Robotics and Mechatronics, German Aerospace Center (DLR)*
²Institut für Informatik,Technische Universität München (TUM)

ABSTRACT

In order to improve the accuracy and reliability of myocontrol (control of prosthetic devices using signals gathered from the human body), novel kinds of sensors able to detect muscular activity are being explored. In particular, Optical Myography (OMG) consists of optically tracking and decoding the deformations happening at the surface of the body whenever muscles are activated. OMG potentially requires no devices to be worn, but since it is an advanced problem of computer vision, it incurs a number of other drawbacks, e.g., changing illumination, identification of markers, frame tear and drop. In this work we propose an improvement to OMG as it has been recently introduced, namely we relax the need of precise positioning and orientation of the markers on the body surface. The small size of the markers and their curvature while adhering to the surface of the forearm can lead to missed detections and misdetections in their orientation; here we rather detect the deformations by applying a Convolutional Neural Network to the region of interest around the feature source segmented, from the forearm. The classification-based approach yields results similar to those obtained by other classification based modalities, reaching accuracies in the range of 96.21% to 99.30% when performed on 10 intact subjects.

INTRODUCTION

Recently introduced in the scientific community of assistive robotics, Optical Myography (OMG) is a novel way of non-invasively detecting the muscular activity [1]. The main idea is simple: muscle activation, for instance when flexing the index finger, induces a quite precise deformation in the forearm due to the enlargement of the muscle belly, and the consequent shifting of the adjacent musculoskeletal structures. This phenomenon is currently being exploited by the techniques called Force Myography (FMG) and Tactile Myography (TMG); in these cases, the aforementioned deformations are detected via pressure / force sensors, used in small numbers (FMG) or in a high-resolution array (TMG). The results are highly promising [2] [3]. In OMG, such deformations are captured using optical recognition alone, that is, by "looking" at the forearm. Whenever the fingers flex, or the wrist rotates and/or extends, small changes in the forearm's volume and shape appear to the trained eye and become apparent if markers are applied to the surface of the forearm; they can be linearly related to the required muscle activations [1] [4]. This possibility leads to applications in, e.g., advanced upper-limb prosthetics: a camera aimed at the stump of an amputee could be able to control the device, either in the real world or in an Augmented / Virtual Reality setup. The main advantage of OMG is probably that it is the "ultimate" non-invasive human-machine interface for the disabled, since it requires no equipment to be placed on the forearm / stump (in principle even markers can be avoided, if the algorithm is smart enough). On the other hand, it naturally suffers of the problems commonly associated to computer vision: dependence on the illumination and focus, loss of precision due to the varying distance between camera and subject, occlusions, missed and/or misdetection of the markers, and surface features of the forearm. In this work we propose an advancement to [1], in which AprilTags [4] [5] were used and tracked by a camera during movement of the fingers. In a new experiment, we rather used a plain sticker, whose deformations was observed by a single camera and then passed to a Convolutional Neural Network (CNN). With this approach, the need for a fixed relation of the camera to the arm is mitigated and the camera can be worn at the arm. Our experimental results show that the classification of five different finger poses are in the range of 96.21% to 99.3% and therefore on par with state of the art methods like surface EMG, Ultrasound or Force Myography.

Related work

Muscle activity exists even after the amputation of the hand [6] [7]. Such activity can be non-invasively detected in a number of ways: through the electrical activity of the motor units (surface electromyography or sEMG [8]), the deformations of the involved body parts (force or tactile myography [2]), listening to the vibrations induced by the muscle motion (mechanomyography [9]), and so on. We hereby concentrate on another modality, termed Optical Myography (OMG), which estimates finger poses by optically observing deformations on the surface of the forearm. This is performed by mounting a web-camera to a fixed set-up frame and by strapping the subject's forearm to it. By preventing the forearm to move relative to the camera, solely its muscle deformations can be detected with the help of fiducial markers such as AprilTags [4]. The 6-D information (translation and rotation) collected from each of these tags are processed and regressed to four different finger poses (trained independent of one another). The small size of the markers can lead to its improper identification. The curvature when stuck to the surface of the forearm also leads to misdetections in orientation. These factors magnify once the arm is released from the set-up in attempt to carry out practical tasks.

EXPERIMENT DESCRIPTION

Setup Description

An elastic band is used to strap a camera onto the forearm. The images are recorded at a resolution of 640x480 pixels at a frame rate of 25 frames per second. Artificial lighting (from LED lights) is used in these experiments to achieve uniformity in the experiments and to avoid changes in illumination caused by natural light. Motion blur was also physically suppressed to a certain extent by using a velcro band around the forearm and the camera to prevent upwards vertical movement. The images are pre-processed to obtain the ROI using the computer vision library OpenCV and then passed to a CNN, which is implemented using the TensorFlow software library.

Figure 1: The experiment setup, where a subject is following the stimuli on the monitor with a sticker stuck on to the left forearm and a camera attached to the arm to capture its deformations

Participants

The participants chosen for the experiment were people with all their fingers intact. Each subject was asked to use around 80% of their maximum force while following a stimulus signal displayed by a virtual hand on a computer screen. The camera was strapped on the subject's arm to simulate an attachment to the base of an active hand prosthesis as shown in Figure 2.

There were two subjects with the dominant hand being their left, while the others were right handed. The average age of the participants (three female and 7 male) is 26.2 ±3.65 years. A plain sticker was stuck on to the anterior side of each subject's arm. Once prepared for the experiment, the subject was asked to place the forearm (freely) on a plain

Figure 2: (left) The sticker attached to the forearm of a subject; (right) segmentation of the sticker

surface. The subject was then shown both the graphical interface used for recording and a virtual 3-D hand model presenting the stimulus signal to be followed. The stimuli used are thumb flexion, thumb abduction (rotation), index flexion, combo flexion (combination of the little, ring and middle finger) and rest (all fingers relaxed), repeated 10 times at equal intervals. The experiments were approved by the Ethical Committee of the DLR and all subjects gave written consent.

Image processing

By using a plain sticker on the forearm, the sticker's bounding box can be used as a region of interest (ROI) and its deformations and slight changes in position can be used as characteristics for the CNN to distinguish between the finger poses. This requires a segmentation of the sticker in each camera image from the background. In order to segment the sticker, the RGB colour space is transformed into a three channel log opponent chromacity (LO) space, a method common in skin segmentation algorithms [10] [11]. In order to cope with intensity issues such as glare, the intensity channel (I channel) is subtracted from the LO-R_g channel and this new channel is used as the base for the rest of the image processing upon normalization.

A fixed ROI is set for the first frame in order to remove unnecessary background. A median filter smoothens the image before an adaptive threshold is used to segment the boundaries. Morphological operations are used in cases where the sticker's boundary merges with the forearm's or the sleeve's due to its placement. The contours of the sticker are then extracted and enclosed within a rectangular bounding box. Necessary conditions are imposed on the bounding box to filter the contours of the sticker. Once the estimated contour and bounding box of the sticker is verified by the user in the first frame (the remaining being automated), a mask of the contour segments only the sticker and sets the regions outside the contour to a pixel value of 0 (black).

Classification method

The network is a simple CNN consisting of two convolution layers with exponential linear unit (ELU) activation and a fully connected layer. Each convolution layer consists of 16 filters. The filters in the first layer have a relatively large size of 11x11 pixels while the second layer's filters are of size 5x5 pixels. The images sent as input to the CNN are greyscale images of size 130x130 pixels. Image sizes are restricted to be smaller than the down-scaled image to preserve information. About nearly a second (800 ms) of delay is imposed to adapt for the reaction time of the subject. Intermediate pose data are also not considered as an input in order to make it a purely classification problem. The data sent in for training are shuffled in a pseudorandom process so that each of the stimuli is uniformly distributed throughout the training set. The training set is then split into batches of 70. The predictions are estimated by optimizing the parameters of the CNN using stochastic gradient descent which minimizes the loss over 20 epochs. The initial learning rate is 0.001; which is then decreased by 5% every succeeding epoch. The loss function is that of the mean of the sparse softmax cross-entropy of the output of the final fully connected layer (the logits). The accuracy is calculated by finding the argmax of the logits and comparing it to the true labels.

"Leave-one-repetition-out" cross-validation was used to evaluate subject accuracy, that is, for each subject and repetition we trained the CNN on nine repetitions and tested on the selected one. We averaged out all results across subjects to yield a global result.

RESULTS

Figure 3 shows the overall accuracy, i.e. the percentage of correctly classified cases versus all observed cases, averaged over all subjects as well as overall precision, i.e. the true positives versus all positives. The mean and standard deviation cannot be taken as the best value over the repetitions since the overall (intra-subject) sample space is small (10 repetitions yielding 10 separate tests). Thus the median and interquartile range (IQR) depict an evaluation closer to that of the true performance of the model. The confusion matrix of the global median is then used to obtain the global accuracy and precision. The final results are displayed in percentage (after normalization).

As a comparison with existing literature, we show in Figure 4 that OMG using CNN (OMG_CNN) performs on par with the other classification-based modalities. We use the results obtained during similar experiments published in the following papers:

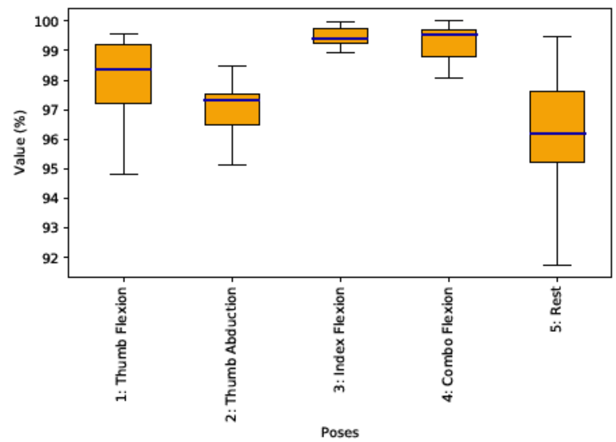

Figure 3: Boxplot of classification accuracy of each pose

Naik et al. [12], Table II, III(A) and III(B) (labelled sEMG in our Figures): Five transradial amputees were engaged in performing 11 finger poses, which were detected using two proposed sEMG configuration which they considered optimal. We take the average of the little, ring, middle, pointer and thumb extension, and the little, ring and middle finger flexion as the rest pose and combo flexion respectively. The remaining poses were the same as in this work.

Cho et al. [2], Table 2 and Figure A1 (FMG): Four transradial amputees performed five trials with 11 different grip gestures. The key grip performed in their study was assumed similar to the thumb flexion in our work, the mouse grip was almost the same as the thumb abduction, the precision open is assumed similar to the index flexion and the finger point and relaxed hand were the same as the combo flexion and rest position respectively. The final confusion matrix drawn from the four subjects' was that of their mean as used by the OMG method between the subjects.

Sikdar et al. [13], Table I (SMG): Ten healthy volunteers performed individual finger flexions. The combo flexion was derived by taking the average of the performance by the little, ring and middle finger flexion, while the remaining were the same as in this work. The rest and the thumb abduction could not be gathered for the comparison.

Consider the bar plots in Figure 4. The index and combo flexion are in fact as high as achieved when using sEMG. One can note that the thumb flexion, thumb abduction and rest position has a lower precision (Figure 4b). This can be explained by the low variation between the three poses, which causes a higher risk of false positives.

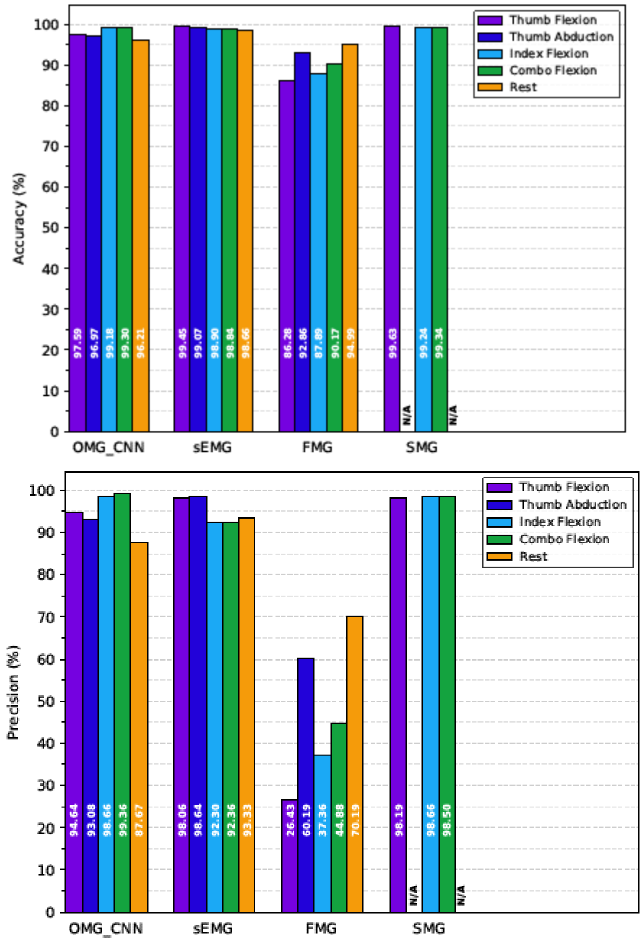

Figure 4: Modality-wise comparison of accuracy (top) and precision (bottom) in percentage between the various finger poses

CONCLUSIONS

In this paper we have proposed a simple improvement to optical myography (OMG), namely, the usage of a single undifferentiated marker ("sticker") instead of several AprilTags, as it was done in [1]. We have demonstrated that such a simple arrangement is enough to obtain classification results, for several movements of the fingers, which are comparable to those already obtained in literature using sEMG, ultrasound imaging and force myography. A Convolutional Neural Network seems to be a good option to take advantage of the image-like nature of the sticker and its deformation due to muscular activity. The main challenges to be faced here are the morphological operations used to disjoin boundaries intersecting with the sticker's during contour extraction. It is interesting to note that thumb movements are not easily distinguishable from each other and from the rest pose, probably due to the fact that the muscles, which control the thumb are deeper and therefore harder to detect on the skin surface using an optical camera.

OMG is, of course, susceptible to all well-known pitfalls of computer vision: motion blur, varying illumination and occlusion(s). This is going to be the main line of future research, especially as we will lift the assumption of the forearm being fixed in a specific spatial position.

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RTM-PDCP LINKAGE PLATFORM MULTI-MODAL SENSOR CONTROL OF A POWERED 2-DOF WRIST AND HAND

[1]Masaki Shibuya, [2]Kengo Ohnishi and [3]Isamu Kajitani

[1]Graduate School of Tokyo Denki University
[2]Tokyo Denki University
[3]National Institute of Advanced Industrial Science and Technology

ABSTRACT

Highly functional multi-DOF upper limb prostheses are available to the users, though problems remain on the control strategies' operation load. We proposed Multimodal Sensor Control, MSC, which integrates myoelectric signal and forearm posture signal to operate the prosthetic hand and wrist. Experiments comparing MSC and locked-wrist myoelectric control showed that compensatory shoulder motion can be reduced with MSC, yet only on specific conditions.

To augment the MSC to variety of daily activities with least operating burden, we propose to combine environmental information to the motion signal, e.g. myoelectric and forearm posture, since hand operation is selected by the relativity of the grasping object posture and hand orientation. The key is the network for streaming the environmental information to the prosthesis controller. A robotized room for comprehensive support environment for the physically handicapped is an expedient and RT-Middleware, RTM, has a proven strategy. Prosthetic Device Communication Protocol (PDCP) is the best test bed for the prosthetic control, and therefore, the objective of this project is to develop and verify the availability of MSC using environmental information with RTM-PDCP linkage platform. As a proof-of-concept model, wearable tag reader was implemented to demonstrate the operation based on the relativity of the hand and environment. By mounting RFID on the grasping target's surface, tag reader on the hand and an inclination sensor to the working table, the information of the tag ID, inclination angle of the object, approaching motion signal of inertia measurement unit and trigger of the myoelectric signals are combined to presume the grasping direction of the target object and switch the servo control mode and drives the 2 wrist motors to maintain the wrist angle to while grasping the object.

The operation load applying the MSC was verified by conducting experiment of 48 trials. A powered wrist and hand was assembled and donned on the right forearm of 4 non-amputees subjects. Six tasks were selected from the therapeutic battery, Simple Test for Evaluating hand Function (STEF), for evaluation. The operating forearm posture angles and work times of the trial with MSC and conventional myoelectric control were measured. The average work time of MSC was larger but not statistically significant, while the average forearm posture angle range was significantly smaller ($p<0.05$). These results of downgraded forearm posture angle range without prolonged operation time demonstrates that the MSC using environmental information can be operated on RTM-PDCP linkage platform and suppresses compensatory motion.

A NOVEL APPROACH TO VISUALISING UPPER LIMB ACTIVITY IN MYOELECTRIC PROSTHESIS USERS

Alix Chadwell, Laurence Kenney, Malcolm Granat, Sibylle Thies, John Head and Adam Galpin

University of Salford

ABSTRACT

There are reports in the literature suggesting that around 30% of people prescribed myoelectric prostheses reject their device. Further, there is limited and often rather ill-defined data on usage in people who do not reject their prosthesis. To date, the only method of determining real-world prosthesis usage has been through self-report. Self-report is subject to inaccurate and potentially biased recall and provides, at best, average usage data. The lack of high quality data on real world use of prostheses is very surprising, given one of the core purposes of providing a prosthesis is to restore upper limb function in everyday life.

Activity monitoring offers a method to objectively characterise upper limb activity outside of the clinic. Wrist-worn activity monitors, comprising of tri-axial accelerometers, have been used successfully on numerous occasions to assess the upper limb activity of healthy anatomically intact people and, for example, people recovering from a stroke. Despite the obvious potential to quantify upper limb movements, the use of activity monitors for the assessment of people with upper limb absence has been surprisingly underexplored.

Recently we published the first data using activity monitors to quantify myoelectric prosthesis use [1]; however, the data visualisation method was limited in scope. Arbitrary values were introduced for unilateral activity, a natural log of the usage ratio was taken making interpretation difficult, and temporal patterns of prosthesis usage were not considered. Based on methods which have previously been used for the time series visualisation of whole body activity data, we have developed a new method for displaying upper limb activity. This method uses spiral plots with graduated colours to display the percentage contribution of each upper limb to activity over a week long period. Using this method it is possible to quickly identify the level of symmetry in a person's arm activity, and distinguish patterns of activity between users.

Here we present data recorded from trans-radial myoelectric prosthesis users who report to be both satisfied and dissatisfied users of their prostheses. Participants wore Actigraph GT3X+ monitors on both wrists (anatomical and prosthesis) for a seven day period.

Spiral plots allow for quick identification of patterns in prosthesis use throughout the week; this could be beneficial to both clinicians and researchers. Furthermore, there is the future potential with this method to overlay additional data, such as self-reported wear, or activations of the hand.

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A STUDY OF THE REALITY OF MYOELECTRIC PROSTHESES TO INFORM FUTURE RESEARCH

Alix Chadwell, Laurence Kenney, Sibylle Thies, John Head and Adam Galpin

University of Salford

ABSTRACT

The first clinical myoelectric prosthesis was developed in the 1960's. Since this time research has seen a number of developments, including pattern recognition of control signals, implantable electrodes, and multi-articulating hands. Of these developments, only multi-articulating hands are widely commercially available, suggesting a problem with technology translation. Furthermore, around 30% of myoelectric prosthesis users subsequently reject their prostheses [1]. Amongst the key reasons cited for prosthesis rejection are poor control and poor functionality.

One of the reasons for the very poor rate of technology translation is likely to be the focus taken by many of the research groups in this area on using intact participants in the early phases of research [2]. Of particular note, early phase studies have generally not taken account of the issues associated with transducing signals from socket-located electrodes. Saunders noted that variability in signal transduction is likely inherent to the design of current prostheses [3]. Building on this, the authors have identified three factors from the literature which may impact on prosthesis user performance. These are the skill of the user in controlling the muscle signals, the unpredictability of hand response introduced by a poor electrode fit, and the delay in the hand response due to inherent electromechanical delays. A protocol was developed to assess each of these factors to establish their relative impact on prosthesis user functionality and everyday use of their myoelectric prosthesis [4].

For this study we are recruiting a range of prosthesis users encompassing those who are both satisfied and dissatisfied users of their prostheses, both wearers and non-wearers. Consequently multiple centres across the UK have been involved in the study.

Here we will present preliminary results demonstrating how each of the control factors (skill, unpredictability and delay) relate to measures of user functionality (including task success, task duration, gaze patterns, hand aperture patterns, and movement variability) and everyday prosthesis use (assessed using activity monitoring over the course of a week).

Early results suggest that unpredictable device response is a key factor. Users whose hand reacted unexpectedly were also more reliant on visual feedback as to the hand state, and less likely to wear their prosthesis during the week in which they were monitored.

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PATTER RECOGNITION MYOELECTRIC CONTROL CALIBRATION QUALITY FEEDBACK TOOL TO INCREASE FUNCTION

Nathan Brantly[1], Aimee Feuser[1], Frank Cummins[1], Levi Hargrove[1,2], Blair Lock[1]

[1]Coapt, LLC, [2]Shirley Ryan AbilityLab

ABSTRACT

Pattern recognition control for upper-limb myoelectric prostheses is growing in clinical acceptance. Furthermore, powered prostheses are becoming increasingly complex, especially with the growing popularity of multi-articulating hands [1]. With a larger available motion set, requiring a larger number of patterns of muscle activity, complications can arise. Calibration feedback to rate the ability to control each available motion and provide tips for subsequent recalibration is highly beneficial in these cases. Here, a novel algorithm for determining calibration quality is presented.

INTRODUCTION

Since commercialization in late 2013, many individuals with upper limb loss and limb difference have benefitted from pattern recognition control for myoelectric prostheses [2]. Since its initial release, many software tools have been developed and deployed, primarily to assist prosthetists in fitting and properly integrating the pattern recognition control system. For instance, a clinician assistant tool provides an interface for clinicians to communicate directly with a support representative by submitting a package including the most recent set of calibration data for troubleshooting and further analysis; a real-time signals analysis tool alerts users to EMG signals that are possibly unreliable (primarily the result of poor skin-electrode contact or signals that are too high and clipping); and a practice tool provides an environment to further explore pattern recognition control with two games: a posture matching game based on the Target Achievement Control (TAC) test and a proportional control game [3].

All of the aforementioned software-based tools have proven to be invaluable. However, there has remained a need to provide clear and concise feedback describing the pattern recognition calibration quality to the user. No clinically focused tool has been developed to rate the quality of calibration data for each available motion. Technical descriptions such as classification error rates have been traditionally used, but they do not adequately explain the underlying cause of poor calibration, such as late contraction initiation, or a contraction performed in the incorrect sequence. A properly designed tool can automatically analyse calibration datasets and provide clear, concise feedback and guidance on ways to modify subsequent recalibration to improve prosthesis control. Without this tool, diagnosing the cause of unsatisfactory prosthesis control, with the exception of EMG signal quality issues such as electrode lift-off or saturation, is difficult, leaving clinicians and users to make educated guesses as to why control may not be as expected. Furthermore, users are left to an unstructured and highly experimental approach to determine preferred calibration techniques. This sometimes requires iterating through a calibrate-and-check-control process multiple times before identifying a user-specific method to achieve satisfactory prosthesis control. Some users may feel uncomfortable with such an unstructured approach to finding a preferred calibration technique, possibly becoming fatigued from the experimentation process or even possibly becoming frustrated with suboptimal control. A software feedback tool should considerably reduce the time necessary to achieve high-functioning control.

CURRENT APPROACH

Improving the quality and character of muscle contraction patterns presented during calibration of a pattern recognition control system improves the capability to form patterns that can be accurately recognized by the pattern recognition system. Preliminary evidence suggests that this also leads to improved functional control of the prosthesis over a multiple week home-trial [4, 5].

Some common approaches to improve the richness of EMG signal pattern information, such as varying contraction strength or increasing the differentiation between contraction patterns, can make a significant difference in prosthesis control and thereby greatly reduce frustration as well as the support burden on clinicians [6, 7].

The current commercial pattern recognition myoelectric control system uses an open-ended calibration format where the only mechanism to validate the control is to attempt to control the movements of the prosthesis. This is an ad-hoc approach that can be discouraging when the resulting control is suboptimal. Without instruction feedback, as an example, users often increase muscle contraction intensity in the case of unsatisfactory control, further obscuring the ability of the controller to decipher intended movements. Providing

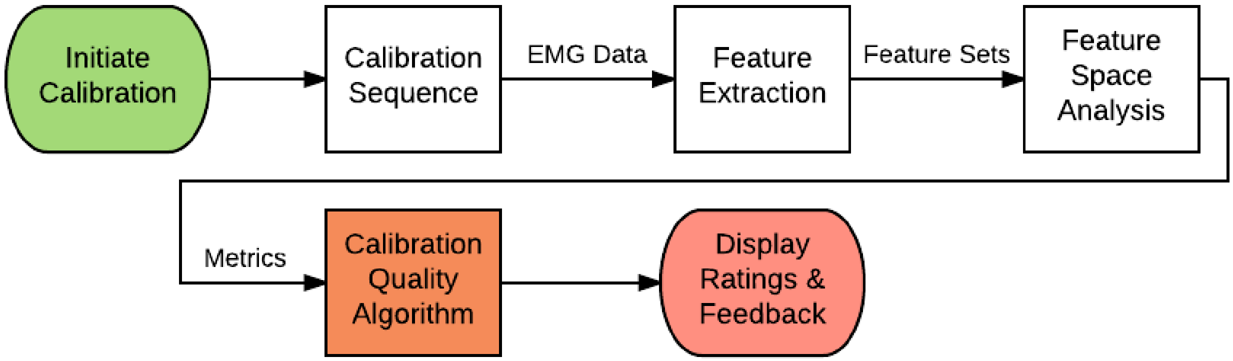

Figure 1: Calibration quality software tool flowchart

assistive feedback can positively reinforce the user experience with pattern recognition as users can be informed of motions that are likely to be suboptimal, the likely underlying causes, and guidance on corrective actions. Furthermore, without a mechanism for feedback, it is not always clear to pattern recognition users in their own environment when it is appropriate to contact their clinician for controls assistance if they are struggling.

CALIBRATION QUALITY SOFTWARE TOOL

Background

The need for a calibration quality software tool has been identified through various forms of qualitative feedback from clinicians who have fit pattern recognition control systems and current users. Unsatisfactory prosthesis control has caused users to contact their prosthetist or support representative for assistance and guidance for control improvement. Providing greater autonomy in identifying ways to improve prosthesis control serves to increase prosthesis function and acceptance.

Purpose

The purpose of the calibration quality software tool is to inform the user of the pattern recognition control system in ways that promote both early-stage controls learning and continued functional growth. The intent of the tool is to analyse each available motion and provide clear, concise feedback messages to instruct users in ways to perform subsequent recalibration differently to improve prosthesis control.

Description

There are 3 major aspects of the calibration quality software tool: 1) the underlying algorithmic processes to evaluate character and quality of users' pattern recognition EMG data; 2) the determination and presentation of the quality rating for each functional motion class; and, 3) the autonomous formation of the instructional feedback messaging to the user (hints for improvement). Figure 1 shows a high level flowchart of the calibration quality input, data processing, algorithm, and output to the software user interface.

Calibration Quality Assessment

The calibration quality tool algorithm primarily analyses users' EMG data after the data has been represented in feature space (Figure 1) – a critical component of pattern recognition, whereby EMG data is modelled in a lower-dimensionality subspace. Two Mahalanobis distance-based metrics, Separability Index (SI) and Repeatability Index (RI), are computed using the feature set data of each available class (equations from Kim et al. 2016) [8-10].

The Mahalanobis distance between two class feature sets is computed as

$$D_M(X,Y): \sqrt{(\mu_X-\mu_Y)^T\tilde{S}^{-1}(\mu_X-\mu_Y)} \tag{1}$$

where μ_X and μ_Y are the means of class X and Y, and $\tilde{S}$ is the weighted covariance matrix between class X and Y, which is computed as

$$\tilde{S}=\frac{n_X}{N}S_X+\frac{n_Y}{N}S_Y \tag{2}$$

where n_X and n_Y are the number of feature sets of class X and Y, and N is the total number of feature sets across both classes. Repeatability Index is computed as

$$RI: \frac{1}{r}\sum_{k=1}^{r}D_M(X_r,X_t) \tag{3}$$

where r is the number of times EMG data was collected during calibration for the class, X_r is the feature sets of the rth class collection, and X_t is the cumulative feature sets across all EMG data collections for the class. A smaller value of RI indicates greater consistency of the muscle contraction pattern. Separability Index is computed as

$$SI(j) = \min_{i=1,\ldots,j-1,j+1,\ldots,N} D_M(X_i, X_j) \quad (4)$$

where the SI of class j is the minimum Mahalanobis distance between motion class j and all other available classes. A larger value of SI indicates greater distance between class j and the nearest neighbouring class.

Quantitative metrics are produced, internal to the algorithm that represent the timing and quantity of EMG contraction data provided by the user during calibration. This detects a multitude of potentially problematic issues in the calibration data and combines the issues, based on level of significance and likelihood to introduce functional difficulties. The resulting metric is a score from 1-5, that is mapped to a 5-star rating system.

Quality Rating

The primary element for reporting calibration quality per functional motion class is a 5-star rating system. This common standard is generally understood where a 5-star rating is of highest quality and 1-star is of lowest quality. This approach is designed to immediately draw user's attention to lower quality ratings and provide natural encouragement for them to review the 'tips' aimed at improving the rating.

Informative Feedback

As potentially problematic issues are detected in the calibration data, resulting in a sub-perfect score, a subset of known messages are presented to the user. These messages are short, informative tips created to correct the detected issue during subsequent recalibrations. Each type of detectable and reportable issue is accompanied by a generic statement within the algorithm and the presentation of the messages is via the software interface immediately following calibration. Examples of the categories of these messages include: "Motion A is very similar to Motion B. Consider X to improve muscle pattern separation"; "No EMG data was detected for Motion C. Consider making stronger contractions for C or be sure to contract when prompted.", and; "Motion D is highly variable. Attempt to discover the most repeatable muscle contraction pattern."

CONCLUSION

Previously, no clinically focused software tool was available to provide feedback to the user to describe the quality of the data used to calibrate a pattern recognition controller. Rather, empirical observation of prosthesis control was required to determine if calibration data was adequate for satisfactory control. The automated procedure presented in this contribution provides a structured framework to provide clinically focused feedback to the user. The resulting "pocket assistant" style tool will be especially beneficial to new users who are attempting to calibrate a pattern recognition for the first time. Initial feedback from beta-users has been positive and the messages provided to the users is continually being expanded.

SIGNIFICANCE

Pattern recognition control of upper-limb prostheses is growing in clinical acceptance. Further improvements to the calibration scheme improve the clinical viability of pattern recognition control in comparison with conventional amplitude-based control approaches.

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DESIGN AND INTEGRATION OF AN INEXPENSIVE WEARABLE TACTOR SYSTEM

[1]Katherine Schoepp, [2]Michael Dawson, [3]Jason Carey and [1]Jacqueline Hebert

[1]Division of Physical Medicine and Rehabilitation, Department of Medicine, Faculty of Medicine and Dentistry, University of Alberta

[2]Research and Technology Department, Glenrose Rehabilitation Hospital and Division of Physical Medicine and Rehabilitation, Department of Medicine, Faculty of Medicine and Dentistry, University of Alberta

[3]Department of Mechanical Engineering, Faculty of Engineering, University of Alberta

ABSTRACT

Commercial myoelectric prostheses do not provide sensory feedback to the user; developing an inexpensive feedback system that can be easily retrofit onto existing prosthetic components may reduce barriers to clinical translation and testing. We describe the development of an inexpensive and wearable tactor-integrated prosthesis, including the (i) evaluation of sensors that can be retrofit onto existing commercial terminal devices, (ii) design and evaluation of custom mechanotactile tactors, and (iii) design of a custom electronics controller which translates sensor input to tactor output.

Three commercial sensors were evaluated for their ability to instrument individual digits, minimize cost, maximize accuracy, and avoid significant alterations to the prosthetic hand. Evaluated technologies include an FSR (Interlink, FSR 400), a subminiature load cell (Honeywell, FSG020WNPB), and a capacitance sensor (SingleTact, S8-10N). A full-factorial design of experiments was conducted to evaluate sensor responses under different loading conditions including material stiffness, loading rate, sensor contact, and indenter curvature. For low-accuracy applications, the FSR is recommended; for high-accuracy applications, the load cell is recommended where modifications to the prosthetic fingertips are possible, otherwise the SingleTact sensor is recommended.

Two mechanotactile haptic displays were designed; a linear and a cable-driven tactor. Both models use the same servo motor (HiTec, HS-35HD), with a rack and pinion gear system to convert rotational motion to linear, where contact to the residual limb is made via an 8 mm diameter domed head. The cable-driven tactor offers a reduced vertical profile at the tactor head site, however it has a larger overall footprint and draws more current. Tactors can be controlled to set a specific displacement or force, with time delays and output accuracies quantified for each system.

A custom electronic controller was designed to map forces from the sensors on the prosthetic fingertips to the haptic display. The system integrates with the existing prosthetic components and can control up to eight individually mapped tactors, where settings can be adjusted wirelessly. It contains four custom boards in addition to a commercial wireless transmitter (SparkFun, WRL-12580); all boards are contained within a custom electronics enclosure which fits into the forearm of the prosthesis.

The sensors, tactors, and electronics were integrated into a commercial prosthetic arm with minimal increases to cost (material cost \$300 plus \$125 per tactor, excluding assembly time) and weight (100 g plus < 50 g per tactor). Evaluation with an amputee participant will be discussed along with limitations and suggestions for improvements.

ROBUSTNESS OF REGRESSION-BASED MYOELECTRIC CONTROL IN A CLINICAL SETTING

Janne M. Hahne[1,*], Meike Schweisfurth[1], Dario Farina[2,1]

[1] Neurorehabilitation Systems Research Group, Department of Trauma Surgery, Orthopedic Surgery and Plastic Surgery, University Medical Center Göttingen, Germany
[2] Department of Bioengineering, Imperial College London, London, UK

*Correspondence to: janne.hahne@bccn.uni-goetttingen.de

ABSTRACT

A miniaturized low-cost, low-power embedded system for regression-based simultaneous and proportional myoelectric control of a hand prosthesis with two degrees of freedom is presented. In a case study on one subject with transradial amputation, this system outperformed two commonly used conventional control techniques. Furthermore, the robustness of the approach against changing arm position and across sessions without retraining the regression model was demonstrated.

INTRODUCTION

Myoelectric signals are commonly used to control electrically powered hand prostheses [1]. Recent developments in mechatronics have led to a number of commercially available dexterous prosthetic hands with many degrees of freedom (DOFs) including individually actuated fingers [2]. However, the progresses in control are behind these developments. Most prostheses are still controlled with very simple techniques, based on two EMG signals from antagonistic muscles of the residual limb. These techniques allow for proportionally controlling only one DOF at a time. Cumbersome heuristics, such as selecting the active DOF by a co-contraction or based on the slope of the EMG onset, are used to control more DOFs sequentially.

Multiple functions can be controlled by using machine learning techniques. In particular, classification-based approaches have been investigated for many years [3]–[5]. Basic classification approaches offer only simple on/off control and a sequential activation of DOFs, but are often extended to include proportional control [6] and simultaneous activation of multiple functions [7]. Recently, regression-based control approaches (RC) have gained increasing interest. With these approaches, simultaneous and proportional control of multiple DOFs is possible [8], [9].

However, the impact of machine-learning based control on clinical practice has been limited. So far only one company offers a classification-based control system as an add-on for prosthetic hands [10]. One reason for the limited impact is related to robustness problems of most machine-learning approaches under real-world conditions. In daily use, factors such as altered arm positon [11], small electrode shifts [12], changing skin conditions [13] and time between training and use [14] can influence the signals and degrade the performance significantly.

For this case study, we implemented a linear-regression based control of two degrees of freedom on a miniaturized embedded system that fits into a prosthetic socket and allows for evaluating the control in real-world conditions. The performance was tested with the standardized clothespin relocation test, in varying arm positions and across different days without retraining the linear mapping model. A comparison with two conventional control techniques was also performed.

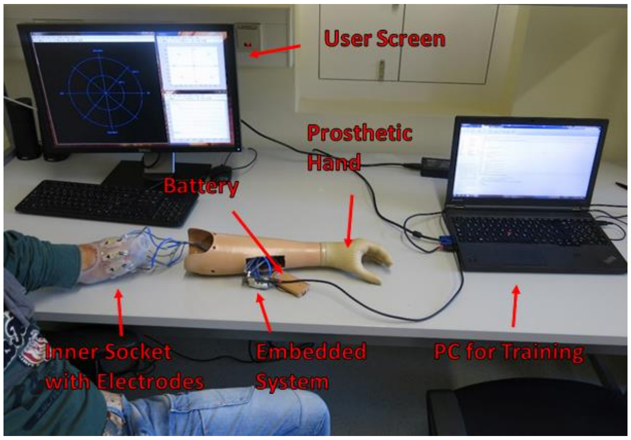

Figure 1: Setup for PC-supported training of user with regression algorithm.

METHODS

Regression-Based Control on Embedded System

The experimental setup (Fig. 1) consisted of eight conventional bipolar electrode modules (Otto Bock, 13E200), a conventional prosthetic hand (Otto Bock, DMC Hand) in combination with a rotation unit (Otto Bock, Electric Wrist Rotator), a custom controller, and a PC.

To implement a portable, miniaturized solution for simultaneous and proportional control of the two prosthetic DOFs, an embedded system was developed (Fig. 2). It consists of an ATMEL ATXMEGA32-A4U 8-bit microcontroller (MC) clocked at 32 MHz with the integrated calibrated RC-oscillator. This MC has an integrated A/D-converter with analog multiplexer that was used to sample the eight EMG signals. As the electrode modules already include temporal filters, rectification, and low-pass-filtering, no additional analog filters were required and a sampling rate of 25 Hz was sufficient.

To control the grasping function of the prosthesis, two electrode signals were emulated by generating analog electrode outputs with the integrated PWM signal generators and external passive RC-low-pass for smoothening. To generate the high currents required for directly driving the motor of the rotation unit, a motor driver was used (ON Semiconductor, LV8548MC). The 3.3 V voltage supply required for the MC was generated from the prosthesis 7.2 V battery by a linear voltage regulator (Texas Instruments, TPS7233).

The implemented firmware provided two modes. For signal inspection and training (see below), it could be connected via USB to a PC and used as a data-acquisition device. The visualization and control algorithm were in this case executed on the PC and control signals could be sent back to the MC to control the prosthesis in real-time. Once the training was finished, the learned regression model could be uploaded to the MC and permanently stored in the integrated EEPROM. Then the system could be disconnected from the PC and used in autonomous mode (Fig. 2 green block), where the eight EMG signals were directly mapped into 2-DOF control signals.

The mapping from EMG envelopes into simultaneous and proportional 2-DOF control signals was performed with linear mapping:

$$\hat{\boldsymbol{y}} = \boldsymbol{W}^{\boldsymbol{T}}\boldsymbol{x} \tag{1}$$

$$\boldsymbol{W} = (\boldsymbol{X}\boldsymbol{X}^{\boldsymbol{T}})^{-1}\boldsymbol{X}\boldsymbol{Y}^{\boldsymbol{T}} \tag{2}$$

where $\boldsymbol{x}$ is a vector with the eight EMG envelopes, $\hat{\boldsymbol{y}}$ the two-dimensional control output, $\boldsymbol{W}$ an $< 8 \times 2 >$ transformation matrix and $\boldsymbol{X}$ and $\boldsymbol{Y}$ matrices with training data and labels.

To suppress unintended motions, activation thresholds where applied for each of the four prosthetic functions at which the prosthesis would start actuating with lowest speed. Upper thresholds were defined at a comfortably reachable maximal regression output for each prosthetic function, which were mapped to the maximal prosthetic speed. The pre-defined values for the activation and upper thresholds were 0.1 and 1. They were adjusted before the experimental evaluation to optimize controllability.

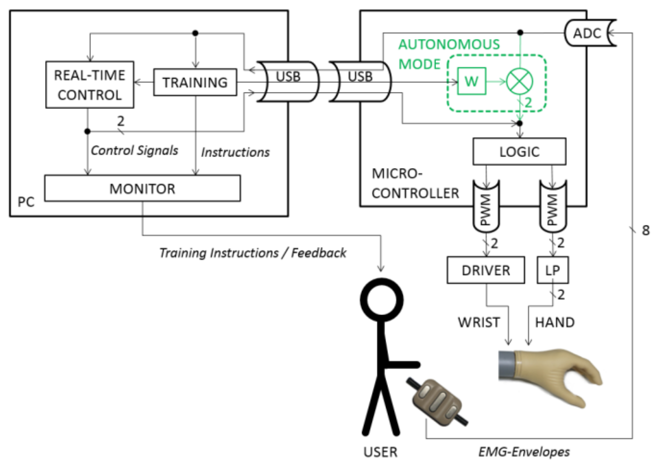

Figure 2: Block diagram of the regression-based myoelectric control system

Training of user and algorithm

For a successful application of machine-learning based myoelectric control systems, both the user and the algorithm need to be trained. The training protocol was based on the co-adaptive learning paradigm developed in [15]. First, the user performed various phantom-limb movements. Out of these, four suitable contraction patterns were selected based on visual inspection of the EMG signals. Then, three runs of calibration data were recorded in neutral arm position, in which the subject followed predefined trajectories that consisted of non-combined movements only and were presented on the user screen as visual cues. An initial linear mapping model was generated based on the least mean-squares solution (Eq. 2). It has been shown that this approach allows for a generalization from non-combined to combined motions [16]. This model was used for real-time control of a cursor within a two-dimensional coordinate-system in position-control mode. The user was given some time to familiarize with the control. Supported by a small computer game in which the user had to catch circular targets in 2D, he was trained and evaluated in the execution of non-combined and combined motions. In the next step, the 2D game was repeated while the linear mapping model was adapted using recursive least squares, so that both the user and the model could improve concurrently.

After a satisfactory mapping model was reached and the entire 2D unity-circle could be firmly accessed by the user, the model was uploaded to the controller and used for prosthesis control.

Experimental Evaluation

The performance was evaluated with the standardized clothespin relocation test that involves both prosthetic functions (open/close and rotation). In this test, the time for moving three red pins (10 N grip force required) from a horizontal to the vertical bar of the Rolyan Graded Pinch Exerciser is measured. As both classification- and regression-based control approaches can be influenced by the position of the arm ([17], [18]), the test was conducted in three different arm positions (arm down, half up, arm up; Fig. 3). Also the time between training and evaluation as well as between donning and doffing of the electrodes can negatively impact the performance [14]. Therefore, the regression-based control was evaluated on two different days, using the model trained in the first day.

As a comparison to the RC, two conventional control systems based on two bipolar electrodes located on the extensors and flexors of the residual forearm were evaluated (using the conventional Otto Bock Myorotonic controller). Co-contraction control (CC) consists of a state machine where a short contraction of both muscle groups triggers a switch of the active DOF. In slope control (SC), the active function is selected based on the slope of the EMG envelope when the contraction is initiated. Slowly increasing EMG-amplitudes are mapped into open/close of the prosthesis, and quickly raising signals into pronation/supination of the prosthetic wrist [1].

This case study was conducted on one male subject with transradial amputation (56 years old, 35 years after amputation). The study was approved by the ethics committee of the University of Göttingen and informed consent was obtained from the participant. A conventional prosthetic socket was constructed that integrated eight equally-spaced electrodes at the location of largest diameter of the residual limb.

Figure 3: Evaluation in the clothespin relocation test, executed in three different arm positions

RESULTS

The completion times needed in the clothespin test are shown in Fig. 4. The proposed regression-based simultaneous and proportional control outperformed the two conventional control techniques CC and SC. The time for completing the task with RC was approximately half the time needed with CC. SC performed better than CC but could not reach the performance of RC. On the first day there was an improvement within the first ten trials of RC, most likely due to learning effects of the user. Remarkably, the arm-position did not impact the performance of RC nor of the conventional control techniques. Even when RC was evaluated on the 2nd day with the regression model of the first day, no degradation in performance occurred (see Fig. 4, in magenta).

With RC, one pin was dropped in trial 2 or the first day (arm position down). With SC, one pin in run 23 (arm position half up) and one in run 26 (arm position up) were dropped. In CC, no pins were dropped.

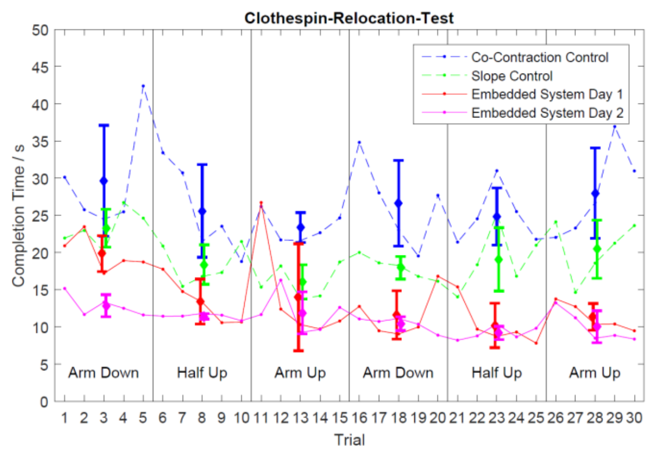

Figure 4: Performance of the regression-based control versus the two conventional control systems, evaluated in the clothespin relocation tests in varying arm-positions.

DISCUSSION AND CONCLUSION

We presented a miniaturized controller for autonomous simultaneous and proportional control based on a linear mapping. In a case study with one transradially-amputated subject, the system was evaluated in a practically relevant task. By varying the arm position and testing the co-adaptively-trained linear mapping model of the first day on a second day, the robustness against three important factors of non-stationarity was demonstrated (arm position, donning/doffing the socket and time between training and application).

The proposed approach outperformed two commonly-used conventional control strategies. The advantage over CC is that the user does not need to perform a time-consuming co-contraction for switching the active function. SC is also sub-optimal since the slope can only be detected in the onset of the contraction and the user needs to relax for a short moment before using another function; in the proposed RC, no break between grasping and rotation is required. A strong advantage over both CC and SC is that even simultaneous motions are possible. This leads to very natural and fluent motion patterns and a fast task execution. In fact, it was observed that the subject made use of simultaneous motions and initiated the rotation already while releasing a pin.

As the current study is limited to a single subject, a follow-up study with a larger number of subjects and over a longer period is still needed to prove clinical feasibility. If successful, the proposed technique has the potential for a clinical transfer in the near future.

ACKNOWLEDGEMENTS

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NEURAL INTERFACE TECHNOLOGY TO RESTORE NATURAL SENSATION IN LOWER-LIMB AMPUTEES

Hamid Charkhkar[1,2], Courtney Shell[2,3], Paul Marasco[2,3], Dustin Tyler[1,2], Ronald Triolo[1,2]

[1]*Department of Biomedical Engineering, Case Western Reserve University, 10900 Euclid Avenue, Cleveland, OH 44106, USA*

[2]*Louis Stokes Cleveland Veteran Affairs Medical Center, 10701 East Boulevard, Cleveland, OH, 44106, USA*

[3]*Department of Biomedical Engineering, Lerner Research Institute, Cleveland Clinic, 9500 Euclid Avenue, ND20, Cleveland, OH, 44195, USA*

ABSTRACT

Sensory input in lower limb amputees is critically important to maintaining balance, preventing falls, negotiating uneven terrain, responding to unexpected perturbations, and developing the confidence required for societal participation and public interactions in unfamiliar environments. Despite noteworthy advances in robotic prostheses for lower limb amputees, such as microprocessor knees and powered ankles, natural somatosensory feedback from the lost limb has not yet been incorporated in current prosthetic technologies.

To compensate for this lack of sensation, amputees rely on visual monitoring of their prosthetic limbs, and often increase loads applied to their intact limbs during standing and walking, putting them at risk for long-term damage. Although there have been numerous attempts to provide sensory feedback via tactile or electro-cutaneous sensory substitution, nothing to date has successfully restored natural sensation that is perceived immediately and directly as coming from the missing limb.

In this work, we report eliciting somatic sensation with neural stimulation in two transtibial amputees. The participants received high-density, flexible, 16-contact nerve cuff electrodes for the selective activation of sensory fascicles in the nerves of the posterior thigh above the knee. In the first subject, the cuff electrodes were implanted on the sciatic nerve just above, and on the tibial and fibular nerves just below the bifurcation. The second subject had two cuff electrodes implanted 3cm apart on the sciatic nerve and one on the tibial nerve. Multiple cuff electrodes were deployed to maximize spatial resolution and increase the likelihood of isolating the desired sensory axons, as well as to explore the degree of selectivity and overlap in responses produced from distal and proximal locations on the nerves. Electrical pulses at safe levels were delivered to the nerves by an external stimulator via percutaneous leads attached to the cuff electrodes.

The neural stimulation was perceived by participants in the study as sensation originating from the missing limb. We quantitatively and qualitatively ascertained the quality, intensity, and modality (pressure, touch, and proprioception) as well as the location of the perceived sensation. Stimulation through individual contacts within the nerve cuffs evoked sensations of various modalities and at discrete locations referred to the missing toes, foot and ankle, as well as in the residual limb. About 60% of 48 contacts in the cuffs produced perceptions 3 months post-implant in response to electrical stimulation.

Based on our findings, the high-density cuff technology is suitable in restoring natural sensation to lower limb amputees.

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SIMULTANEOUS CONTROL OF A VIRTUAL MULTI-DEGREE OF FREEDOM PROSTHETIC HAND VIA IMPLANTED EMG ELECTRODES

[1]Hendrik Dewald, [1]Platon Lukyanenko and [2]Matthew Williams

[1]Case Western Reserve University
[2]U.S. Department of Veterans Affairs

ABSTRACT

The loss of a hand can have a profound, life-altering impact on an individual's life. Recent advances in dexterous myoelectric hands now afford a device that can come close to replicating the range of grasp shapes possible by the natural hand. However a bottle-neck remains regarding the ability to naturally convey enough information to control the multi-degree of freedom (DOF) hands now available. This work describes the deployment of an intuitive multi-DOF command interface that uses permanently implanted EMG electrodes in the muscles of the residual limb and machine learning techniques to decode user motor intent.

This paper covers the results of the first recipient of such a system. The implanted portion consists of eight bi-polar EMG electrodes surgically placed within the proximal muscles of the wrist and fingers. The leads were tunneled under the skin to the upper arm and exit percutaneously. A cued posture matching task was used to train the system to recognize user intent to control the simultaneous velocity of three degrees of freedom – hand aperture, wrist flex/extension, and wrist rotation. An artificial neural network (ANN) was used to convert key magnitude and frequency-related features of the EMG signals into continuous joint velocity. An ensemble approach was used whereby the coincident features of all eight EMG signals are used as inputs instead of the one-muscle, one-action agonist/antagonist command approach used in more conventional myoelectric prosthetic hands. A virtual reality (VR) posture matching task was used to test system performance. The performance of the intact hand was used as a standard for comparison.

The results of this work have shown that the implanted electrodes offer superior signal to noise ratio and less electrical interaction between electrode pairs than that seen using surface EMG recordings. User signals for specific hand motions have been consistent over several months, allowing for literal plug-and-play operation without the need for regular re-calibration. Compared to the intact hand, the ANN decoded movements exhibited no significant difference ($p > 0.05$) in terms of Trial Time (3.39±0.13 vs.2.82±0.21 s), Overshoot (41±8% vs.48±8), or Success Rate (100%), though they did exhibit a slightly reduced Path Efficiency (57±2 vs. 66±2%) and slower Movement Speed (18±0.5 vs. 22±0.4 %ROM/s).

This work demonstrates the potential benefits of coupling machine learning techniques with implanted ensemble EMG recording. Improvements in signal quality yielding more consistent signals can possibly afford prosthetic hand performance on par with the intact hand.

HIGH DEGREE-OF-FREEDOM CONTROL OF VIRTUAL AND ROBOTIC PROSTHETIC HANDS USING SURFACE EMG

Suzanne Wendelken, Tyler Davis, Christopher Duncan, Jacob Nieveen, David Kluger, David Page, Jake George, Douglas Hutchinson, David Warren and Gregory Clark

University of Utah

ABSTRACT

Dexterous, intuitive, multi degree-of-freedom (DOF) control of a prosthetic hand is a highly sought after feature for next-generation multi-articulated robotic prosthetic hands. Here we present results of ongoing studies in which transradial amputees and intact subjects instrumented with 14-to-22-electrode surface EMG (sEMG) assemblies were able to simultaneously control 6-to-8-DOF of a virtual or robotic hand, and generate novel grasps that were not previously trained.

Subjects were instrumented with up to 22 "wet" sEMG electrodes (Coviden, Mansfield, MA), or a sleeve containing ≥ 14 dry "button" electrodes (Motion Control, Salt Lake City, Utah) placed on the forearm or residual forearm clustered above digit and wrist muscles. Decode calibration data was collected at 1kHz using a bioamplifier (Ripple LLC, Salt Lake City, Utah) while the subjects followed repeated single-DOF movements of a virtual hand (e.g., index finger flexion) and one full-hand grasp movement. Data were filtered with a 15-375Hz bandpass filter. Amplitude of single-ended and software-differenced channel pairs were computed, binned in 33 ms windows, and used as the input to a Kalman filter decode algorithm, capable of position or velocity decoding modes. To further minimize crosstalk between DOFs, experimenter-selected gains and thresholds were applied to the outputs, which were then used to control in real-time a virtual hand in a virtual environment (MuJoCo), or a 6-DOF robotic prosthetic hand (DEKA, Manchester, NH). Individual DOF control was verified by means of a virtual target-touching task, where subjects were instructed to touch and hold single or multiple-DOF targets while holding the non-target DOFs in a neutral position. Multi-DOF, untrained movements and grasp and positions were further evaluated using the robotic or virtual hand during functional tests such as utensil holding and cup pouring.

A transradial amputee, using a 22-wet-electrode sEMG assembly, achieved 8-DOF control in a target-touching task (45/48 successful trials), similar to his performance using an implanted 32-electrode EMG assembly (47/48 successes). An intact subject was capable of 6-DOF control in a target-touching task using a 14-dry-electrode sEMG sleeve (30/30 successes). Subjects were also able to demonstrate the ability to make novel grasps (such as thumb-index pinch) in the virtual environment.

These studies show that simultaneous high-DOF control of a prosthetic hand using sEMG is possible and similar in performance to iEMG assemblies within sessions, although long-term stability has not yet been demonstrated. Our decoding strategies represent a novel and effective alternative to the commonly used "direct control" or nominal classifier strategies.

ONLINE TACTILE MYOGRAPHY FOR SIMULTANEOUS AND PROPORTIONAL HAND AND WRIST MYOCONTROL

Christian Nissler, Mathilde Connan, Markus Nowak and Claudio Castellini
German Aerospace Center (DLR), Institute of Robotics and Mechatronics, Wessling, Germany

ABSTRACT

Tactile myography is a promising method for dexterous myocontrol. It stems from the idea of detecting muscle activity, and hence the desired actions to be performed by a prosthesis, via the muscle deformations induced by said activity, using a tactile sensor on the stump. Tactile sensing is high-resolution force / pressure sensing; such a technique promises to yield a rich flow of information about an amputated subject's intent.

In this work we propose a preliminary comparison between tactile myography and surface electromyography enforcing simultaneous and proportional control during an *online* target-reaching experiment. Six intact subjects and a trans-radial amputee were engaged in repeated hand opening / closing, wrist flexion / extension and wrist pronation / supination, to various degrees of activation. Albeit limited, the results we show indicate that tactile myography enforces an almost uniformly better performance than sEMG.

INTRODUCTION

Dexterous myocontrol is the study of natural control of a dexterous prosthesis by (so far, mostly) upper-limb amputees. By "natural" it is here meant, that such a control should work transparently to the subject, enforcing simultaneous and proportional (s/p) activation of a multi-degree-of-freedom (DoF) prosthetic artefact, directly upon the subject's desire [Jiang et al. (2009)]. Surprisingly, even after 20 years of research, the problem is still open, from a number of points of view. First and foremost, upper-limb prosthetic devices are still heavy, noisy, power-consuming and cumbersome; second, non-invasively or minimally-invasively extracting enough information from the subject's body to drive up to ten DoFs is a challenge; last but definitely not least, enforcing reliability of such a control proves to be hard due to the inherently statistical nature of machine-learning approaches used to enforce it, as well as to the changing nature of the signals yielded by surface electromyography (sEMG). Extensive surveys (e.g., [Micera et al. (2010), Peerdeman et al. (2011), Ison and Artemiadis (2014), Engdahl et al. (2015)]) show that solving these three problems would lead to greater acceptance and more extensive usage of such costly devices.

Among the proposed avenues to solve them, we here focus on *multi-modal sensing* [Jiang et al. (2012), Fang et al. (2015)]; in particular, force myography (FMG) and its high-resolution counterpart, tactile myography (TMG) are showing very promising results. Almost 20 years have now gone by since Kenney and Craelius's seminal works [Kenney et al. (1999), Curcie et al. (2001)] on the detection of stump deformations as an alternative to sEMG [Merletti et al. (2011)]; and the applications are now out in the academic world [Cho et al. (2016), Radmand et al. (2016)]. In particular, TMG has the advantage of providing a more stable signal than sEMG [Connan et al. (2016)] and, due to its high spatial resolution (up to 5mm), a richer image of the underlying muscle activity.

In this specific work we describe an experiment in which TMG was compared as fairly as possible with sEMG, during an online target-reaching task aimed at hand and wrist s/p control. We fitted six intact subjects and a trans-radial amputee with a shape-conformable tactile bracelet, and induced them to reach predetermined graded activations of the hand opening / closing, wrist flexion / extension and wrist pronation / supination; the experiment was then repeated using 20 commercially available sEMG sensors. Using several performance measures, TMG showed superior results with respect to sEMG: it enforced a higher Success Rate (SR), shorter Times to Complete each Task (TCT) and longer Time In the Target area when the tasks would fail (TIT). The results obtained by the amputated subject are quantitatively worse than those obtained by the intact subjects, but he still completed more than twice as many tasks successfully with TMG than with sEMG.

As far as we know, this is the first time that TMG-based full online s/p control of the hand and wrist is enforced; the encouraging results we obtained let us claim that TMG should be used as an alternative to, or as a companion of, sEMG in dealing with dexterous myocontrol.

MATERIALS AND METHODS

Experimental setup

TMG data was gathered using a custom-made shape-conformable tactile bracelet based upon the resistive principle [Kõiva et al. (2015)] consisting of 320 tactile sensors (taxels) distributed on ten rigid submodules evenly distributed around the proximal end of the subject's forearm or stump. For further details about the device, please refer to the above-mentioned paper.

sEMG data were gathered using 20 commercially available myoelectric sensors (*MyoBock 13E200* by Ottobock GmbH), arranged on two bracelets, covering approximately the same surface and location of the subject's forearm as the TMG device did (see Figure 1). The sensors were wirelessly connected to the PC using a custom-built wireless ADC device [Connan et al. (2016)].

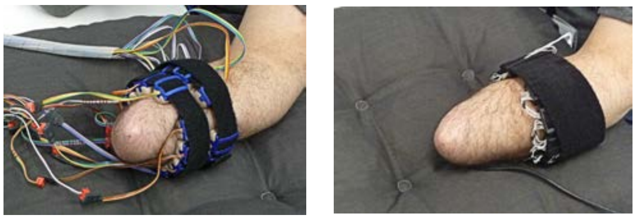

Figure 1: the amputated subject wearing the two sEMG bracelets (left) and the tactile device (right).

To test the approach we used a realistic 3D hand model displayed on a computer screen. Although the model has about 20 DoFs and roughly represents a human hand (including polygon-based 3D rendering and shading), most DoFs were coupled to one another. In the end only three DoFs were considered, namely wrist rotation, wrist flexion / extension and hand opening / closing. More in detail, five specific configurations of the model *(actions)*, namely hand opening / closing, wrist pronation, wrist supination, wrist flexion and wrist extension, were used, each one corresponding to coordinated, graded motions of the three DoFs.

S/p control was enforced using three parallel instances of, in turn, Ridge Regression (RR) applied to the 320 TMG signals and Ridge Regression with Random Fourier Features (RR-RFF) applied to the 20 sEMG signals (both signals were previously mildly low-pass filtered, but no feature extraction was enforced). RR is a well-known linear regression method – essentially least-squares regression plus a regularisation term [Hoerl and Kennard (1970)]. RR-RFF is a non-linear extension to RR, finitely approximating a Gaussian kernel, already successfully employed in myocontrol several times [Gijsberts et al. (2014), Strazzulla et al. (2016)]. Notice that the three DoFs of the model were always operated simultaneously and proportionally, since both RR and RR-RFF are pure regression approaches (i.e., no classification involved).

Subjects and experimental protocol

The experiment was joined by six intact subjects (30.7±7.2yrs old, five males, one female) and one left-hand trans-radial amputated subject (35yrs old male, amputation in 2005, routinely using a *Variplus* hand by Otto Bock GmbH with standard two-electrode control since 2012). All subjects signed an informed consent form; the experiment was performed according to the declaration of Helsinki, and it was previously approved by the DLR Work Safety Committee.

The subjects would comfortably sit in front of the screen displaying two 3D hand models; one of the model would act as a visual stimulus, i.e., the subjects were asked to do what that hand was doing, while the other would show the predicted intended action. The experiment consisted of two identical parts, one performed using the TMG device and one performed using the sEMG sensors. Half of the subjects started with TMG then proceeded to the sEMG part; the order was reversed for the other half. Figure 2 shows an intact subject while performing the experiment.

Figure 2: an intact subject performing the experiment with sEMG sensors. The grey hand is the visual stimulus, while the orange one is the prediction. A smiling face indicates that the current task was accomplished.

Initially each subject performed three repetitions of each required action (plus a "rest" position) while following the visual stimulus; data collected during this phase were used to train the control method at hand (RR for the TMG part, and RR-RFF for the sEMG part); training took not more than 300ms. Subsequently, 30 *tasks* in randomised order were administered to the subjects, as follows: the visual stimulus would perform an action to either full, two-third or one-third activation; the prediction model would then be activated, and the subjects were simply asked to have the prediction model mimic what the stimulus was doing. Intermediate levels of activation were used to determine whether proportional control could actually be achieved, e.g., that the wrist could be flexed at two thirds of the maximum activation. Each task was successful if the subject could match and keep the desired action at the desired activation level for 1.5s; "matching" was defined as remaining within 15% of each target DoF value. If she/he was not able to do so within 15s, the task was declared failed. A visual cue (smiling or sad face) was given as the result of a successful / unsuccessful task.

To evaluate the performance of each method we calculated the ratio of successful tasks over 30 (Success Rate, SR), the time it took the subject to accomplish the successful tasks (Time to Complete a Task, TCT) and the total time spent within the goal for unsuccessful tasks (Time In the Target, TIT). In the latter case (unsuccessful tasks), at this stage we did not differentiate the two sub-cases in which either the required DoF activation could not be reached, or other DoFs would be unwillingly activated at the same time.

The amputated subject was administered exactly the same procedure, executing first the sEMG part.

EXPERIMENTAL RESULTS

Intact subjects. SR was evaluated statistically using a paired Student's t-test – Figure 3 shows the SR comparison. The t-test returned $p = 0.0952$), which means that the difference is not significant ($\alpha = 0.05$). However, the average performance of the tactile bracelet ($75.56\% \pm 21.26\%$) was around 20% better than the performance of the sEMG sensors ($55.56\% \pm 16.42\%$). Furthermore, in case of TCT and TIT the tactile bracelet outperformed the sEMG sensors with $TCT_{TB} = 4.93s \pm 0.95s$, $TCT_{sEMG} = 6.28s \pm 0.58s$, $TIT_{TB} = 1.33s \pm 1.38s$ and $TIT_{sEMG} = 1.91s \pm 0.36s$. These results are summarised in Figure 4.

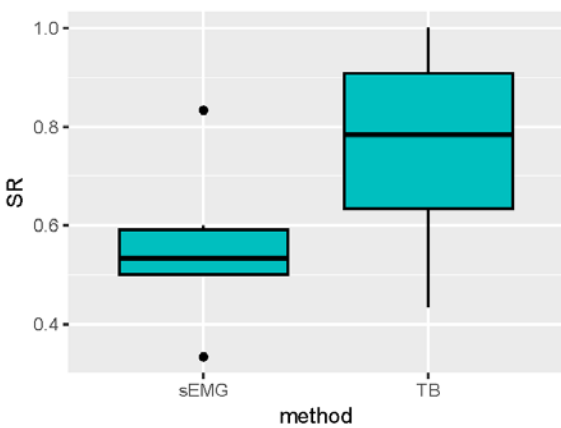

Figure 3: boxplot of the performance comparison between TMG and sEMG sensors.

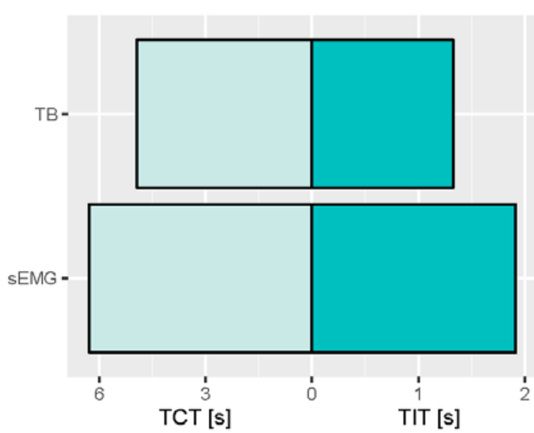

Figure 4: results of the comparison between TMG and sEMG in terms of task completion time (TCT) and Time In the Target (TIT).

Amputated subject. The amputated subject obtained the following results for each method:

sEMG:

$SR = 20\%$
$TCT_{sEMG} = 6.04s \pm 4.37s$
$TIT_{sEMG} = 0.84s \pm 1.48s$

TMG:

$SR = 43.33\%$
$TCT_{TB} = 4.80s \pm 2.38s$
$TIT_{TB} = 0.32s \pm 0.63s$

His success rate is more than double with TMG than with sEMG; as well the required TCTs are on average 20% better (shorter) with TMG. As opposed to this, the TITs obtained with TMG are considerably shorter. (Notice: the larger the TIT, the better.)

DISCUSSION AND CONCLUSIONS

Although preliminary since we tested only six intact subjects and one amputated subject, the experimental results we presented look very promising. For the comparison with TMG (which we enforced using a custom-built device with 320 sensors), we used 20 commercially available sEMG sensors, a very high amount if compared with relevant literature, which potentially poses serious challenges for the embedding in a prosthetic socket. Still, for intact subjects, TMG outperformed sEMG from all points of view considered (SR, TCT and TIT), although statistical significance is still under question (but notice the relatively low number of subjects tested). The amputated person obtained similarly better results with TMG for SR and TCT, but this result must be taken with two important considerations: first, the subject was totally untrained to activate his wrist; second, sEMG was administered first, which might have caused a competitive bias in favour of TMG. (The first remark explains his low overall performance.) Interestingly, his TIT is on average *larger* when using sEMG than TMG; this might indicate that some specific actions were almost unfeasible with TMG, as opposed to sEMG. Further analysis is required to shed light on this issue.

In the only direct reference to a competitor approach we are aware of, namely [Radmand et al. (2016)], a rigid cylindrical encasing fitted with 126 taxels was used to classify eight activation configurations performed by ten intact subjects; body postures were also taken into account by having the subjects perform the tasks in eight different positions in front of them. Since classification was used in this experiment, we cannot offer any direct comparison; their excellent results (classification rates uniformly close to 100%) further indicate the potentiality of TMG.

Lastly, let us remark that in this work linear regression (in the regularised form of Ridge Regression) *directly applied to the mildly filtered tactile values* was sufficient to obtain the results shown. On one hand, this opens up the immediate possibility to embed the whole approach in a prosthetic socket; on the other hand, we will explore in the near future several different sets of features extracted from the tactile image, possibly inspired by image processing, in order to reduce the dimensionality of the input space, and to exploit the reciprocal proximity of the adjacent taxels.

The final proof of feasibility of TMG is obviously to be drawn out of real-life experiments, in which the subject's body posture, the weight of the grasped objects, and the artefacts induced by bumping and accelerations, will need to be taken into account.

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THE CLINICAL APPLICATION OF A MYOELECTRIC TRAINING TOOL FOR UPPER LIMB AMPUTEES

[1]Glyn Murgatroyd, [2]Michael Dawson and [3]Jacqueline Hebert

[1]Glenrose Rehabilitation Hospital, Alberta Health Services
[2]Research and Technology Department, Glenrose Rehabilitation Hospital and Division of Physical Medicine and Rehabilitation, Department of Medicine, Faculty of Medicine and Dentistry, University of Alberta
[3]Division of Physical Medicine and Rehabilitation, Department of Medicine, Faculty of Medicine and Dentistry, University of Alberta

ABSTRACT

The capabilities of commercial myoelectric devices have been steadily improving; however, the process of training and evaluating amputees in using the technology remains challenging especially as the level amputation or complexity of the technology increases. In order to achieve a successful fitting a balance must be struck between providing a device that is as functional as possible yet not so complicated that amputees have difficulty learning or using it. To help find this balance, an interdisciplinary team from the University of Alberta and Glenrose Rehabilitation Hospital (GRH) developed a Myoelectric Training Tool (MTT) that has similar functionality to commercial prostheses. The MTT includes a desktop robotic arm with 5 degrees of freedom, an electromyography (EMG) acquisition system, an embedded controller, and a laptop with a graphical user interface for fine tuning EMG parameters such as gains and thresholds. In 2015 the MTT was successfully translated to the GRH where it has been used for training of muscle control signals. It is also used for potential myoelectric users try the technology and align their expectations to the current state of the technology before fitting them with an actual device. The clinical team use the MTT to assess the number of degrees of freedom the patient can reliably control, to explore control strategy options, and to start training the patient earlier with tasks closer to what they would be able to do with their final prostheses. In this presentation we will describe the occupational training protocol that we have developed for the MTT along with representative case studies. The protocol includes a number of movement and grasping tasks with graded difficulties that are appropriate for training patients with various numbers of muscles sites including those at the transradial or transhumeral level as well as those that have undergone targeted reinnervation surgery. We have developed a method for finding the best EMG sites using the 8 EMG channels available on the MTT. For patients that lack anatomical landmarks a splint or liner material can be used to improve electrode placement consistency as well as provide compression of the electrode into the muscle belly to improve signal output. Future work will focus on further refining the protocol and including more training options for advanced controllers that use pattern recognition.

THE CONTROL BOTTLENECK INDEX: A NOVEL OUTCOME METRIC PROVIDING GENERALIZABLE AND ACTIONABLE ASSESSMENT OF UPPER-LIMB PROSTHETIC SYSTEMS

Dan Blustein, Satinder Gill and Jon Sensinger

Institute of Biomedical Engineering, University of New Brunswick

ABSTRACT

Assessment tools for upper limb prostheses lack generalizability and provide little actionable information to improve outcomes. We have developed the Control Bottleneck Index (CBI), a new outcome measure driven by a computational motor control framework that can provide useful and actionable information regarding system performance. The CBI uses an existing hierarchical Kalman filter model (Blustein & Sensinger 2016, Johnson et al. 2016, Berniker & Kording 2008) to predict system parameters including sensory feedback uncertainty, controller uncertainty, and internal model uncertainty based on human psychophysics results. The CBI tasks include a baseline grasping task, a movement matching task without feedback and a two-alternative forced choice task to determine the just noticeable difference of a perturbation. Here we present this human movement assessment tool and demonstrate its responsiveness to changes in system parameters. The CBI is not task specific but we describe an implementation using a time-constrained grasping task in the Multi-Joint dynamics with Contact (MuJoCo) physics engine with endpoint visual feedback and EMG control (Todorov et al. 2012). We show that when sensory noise and control noise are experimentally adjusted, the CBI can predict changes in the expected underlying motor control model parameters. By inversely calculating model parameters using experimental results, this assessment framework can determine the relative levels of uncertainty of different system components in a human controlling a prosthesis. The results of the CBI can directly inform clinical improvements. For example, if a human performer exhibits a high level of sensory feedback uncertainty, then clinicians can direct improvements to the feedback systems on the prosthesis. The CBI represents a new assessment tool that provides generalizable and actionable information for clinicians working to improve upper limb prosthetic systems.

THE INFLUENCE OF TARGETED MUSCLE REINNERVATION ON PHANTOM LIMB PAIN

[1]Agnes Sturma, [1]Laura Hruby, [2]Martin Diers, [3]Herta Flor, [1]Tanja Stamm, [1]Cosima Prahm, [1]Johannes Mayer and [1]Oskar Aszmann

[1]Medical University of Vienna
[2]Ruhr-University Bochum
[3]Central Institute of Mental Health Mannheim

INTRODUCTION

With an incidence of 50-80% Phantom Limb Pain (PLP) is a big challenge in amputation rehabilitation. Pain medication, psychosocial interventions and novel therapies such as mirror therapy, graded motor imagery, exergames and others were shown to be beneficial in small studies. Another current treatment strategy might be Targeted Muscle Reinnervation (TMR). It has already been shown that TMR can prevent neuroma pain. Additionally, increased intensity levels of PLP in amputees have been associated with cortical remapping in certain brain areas. Here, fMRI studies could show that TMR alters cortical reorganization processes subsequent to amputation. Therefore, it might also decrease the PLP associated with it.

METHODS

To investigate the influence of TMR on PLP a cohort study with two intervention groups and one control group was conducted. While the first intervention group only had a TMR surgery, the second intervention group also received post-surgical rehabilitation and a prosthetic fitting controlled by the re-innervated muscles. Pain levels were assessed before surgical intervention and every 6 months after. Patients were included if they reported mean pain levels of VAS 3 and above, had no nerve injuries and no psychiatric conditions. The study was approved by our Ethics Review Board.

RESULTS

At the current time 7 patients were assigned to the different study groups and have at least completed the one-year-follow-up: 2 patients are in control group, 3 in the first intervention group and 4 in the second intervention group. While the control group reported higher pain levels over time (mean VAS 4.3 in the beginning vs. 5.8 at one-year follow-up), the pain levels of the first intervention group improved from 6.8 to 5 and the pain levels in the second intervention group changed from 5.1 at baseline to 4.0 when using the prosthesis for six months.

DISCUSSION

The results of this study support the previous findings and anecdotal reports that TMR has the potential to relief PLP. As the number of subjects who already completed the study is small, it is not possible to come to any conclusions whether rehabilitation and prosthetic fitting after TMR are reasonable measures to improve this effect. A larger sample size will also allow statistical calculations.

EVALUATION OF CLASSIFIERS PERFORMANCE USING THE MYO ARMBAND

Irene Mendez, Ásgerdur A. Pálsdóttir, Dagbjört H. Eiríksdóttir, Mathias Faulkner, Nicksan Sriranjan, Asim Waris, Ernest N. Kamavuako

® *Department of Health Science and Technology, Aalborg University, Aalborg, Denmark*

ABSTRACT

To provide amputees with intuitive prosthetic control systems, surface electromyography (EMG) has shown promising results in several studies. Myo armband (MYB) is a wireless, ready-to-use technology developed by Thalmic Labs, able to record eight EMG channels with limited frequency bandwidth (<100 Hz). The aim of this study was to evaluate the performance of five classifiers in order to assess the suitability of the MYB to provide reliable accuracy in comparison to the conventional EMG systems (CONV). Eight able-bodied subjects performed nine hand gestures in a crossover acquisition design. Six time-domain features were extracted from the data to evaluate the offline classification error of five classifiers: Linear Discriminant Analysis (LDA), Support Vector Machine (SVM), K-Nearest Neighbors (KNN), Naive Bayes (NB) and Neural Networks (NN). Friedman's test showed no significant difference between CONV and MYB with six channels (P=0.10) with mean classification error of LDA (5.82 ± 3.63% vs. 9.86 ± 8.05%), SVM (9.70 ± 6.02% vs. 11.01 ± 8.79%), KNN (8.30 ± 6.00% vs. 11.12 ± 8.94%), NB (12.48 ± 8.51% vs. 13.77 ± 9.76%) and NN (1.77 ±1.28% vs. 4.64 ± 4.25%) for CONV and MYB, respectively. Although lower classification error was obtained, no significant improvement was found between MYB using eight and six channels (P=0.16).

INTRODUCTION

Surface electromyography (EMG) is a non-invasive technique that records the electrical activity of the muscles during contraction. In the pursuit of intuitive control of multifunctional upper-limb prostheses with several degrees of freedom, numerous studies have applied pattern recognition on EMG signals to identify different movements [1-13]. In the myoelectric control framework, pattern recognition assumes that the signals generated by similar gestures, contain similar features, which are distinguishable from other movements. In the pursuit of improved performance, several studies have compared classification accuracy of various classifiers, such as Support Vector Machine (SVM) [1-5], Linear Discriminant Analysis (LDA) [1-5], K Nearest Neighbor (KNN) [1-5], Bayesian Classifier (BC) [6,7] and Neural Networks (NN) [2-6], among others.

In the recent years, Thalmic Labs has developed the Myo armband (MYB), which is a wireless wearable technology able to record surface EMG. MYB is a multisensory system that records these EMG signals through eight stainless steel electrodes, placed on the forearm, with a maximum sampling frequency of 200Hz. Additionally, it includes a nine-axis inertial measurement unit, haptic feedback and Bluetooth 4.0 communication [8]. Although MYB was initially intended for entertainment, its compact design and intuitiveness have expanded its application into the biomedical engineering field such as in prosthetic hand control [9], or to provide medical image hand-free navigation in a surgical room [10].

From the myoelectric control perspective, the major limitation of the MYB is its restricted maximum sampling frequency. In pattern recognition studies, sampling frequencies above 200Hz have been extensively used [2-5,7] to capture the entire EMG frequency band. Li et al. [11], studied the relationship between sampling frequency and classification accuracy in EMG pattern recognition. Eleven hand motions were sampled at 1 kHz and downsampled up to 100Hz with a 20Hz decrease. Results indicated that classification accuracy decreases with decreased sampling, especially for sampling frequencies below 400Hz.

Therefore, the aim of this study was to evaluate the performance of five classifiers to determine the suitability of MYB for myoelectric control, despite its narrow EMG signal bandwidth (<100Hz). For this purpose, MYB was compared to a full bandwidth EMG acquisition system (CONV) in a crossover study design.

MATERIALS AND METHODS

Data Acquisition

EMG signals were recorded following a crossover study design using both acquisition systems, from eight abled-body subjects (five females/ three males, ages: 19-25 yrs.). The ethical committee of North Jutland approved the experiment. Despite the randomized system order for each subject

Figure 1: Recorded hand movements a) Open hand, b) Closed hand, c) Wrist extension, d) Wrist flexion, e) Pronation, f) Supination, g) Key grip, h) Pinch and i) Rest.

acquisition, the MYB was placed first at 2cm distal to the elbow, to mark the location of six CONV (disposable Ag/AgCl bipolar electrodes) channels. CONV signals were recorded using a custom software at 2kHz sampling frequency and analogically filtered between 10-500Hz. On the other hand, MYB data was sampled at 200Hz. An acquisition software developed in MATLAB using Myo SDK MATLAB Mex Wrapper toolbox [12] was used to record the signals sent via Bluetooth 4.0.

Nine gestures (see Figure 1) of 4s duration, were recorded form each subject in randomized order. Each subject repeated the set five times. To avoid fatigue, subjects had 15s breaks between movements and three minutes' pause between sets.

Data Processing

Due to the difference in the channel number, only six MYB channels (MYB6) were considered for the comparison with CONV. In addition, the same processing steps were also applied to the eight-channel configuration (MYB8) to evaluate the possible classification improvement. We focused only on EMG sensors.

From the 4s movement recordings, only the middle 3s of the contraction were analyzed, to focus on the region with constant contraction force – steady state. [13-15]

The obtained signals were segmented using an overlapping window of 200ms with 50ms increment. Six time domain features [16,17] were extracted from these segments: waveform Length (WL), Mean Absolute Value (MAV), Willison Amplitude (WAMP), Cardinality (CARD), Slope Sign Changes (SSC) and Zero Crossings (ZC). The formulation of the last four features implies a threshold optimization to build a robust feature space. Preliminary work suggests that a threshold of 0.1:1 times the root mean square of the rest signal is required for CARD and WAMP. However, according to [18], no threshold is required for ZC and SSC. Hence, in this study the root mean square of the rest signal was only applied as threshold for CARD and WAMP. Principal component analysis was used to reduce the resulting feature space, preserving 95% of the variance.

Finally, five supervised classifiers were tested: Linear Discriminant Analysis (LDA), Support Vector Machine (SVM), K-Nearest Neighbors (KNN), Naive Bayes (NB) and Neural Networks (NN). The number of neighbors and hidden neurons in KNN and NN were optimized resulting in 1 neighbor and 13 hidden neurons for CONV, 13 neighbors and 14 hidden neurons for MYB6 and 15 neighbors and hidden neurons for MYB8.

To maximized the amount of training data a five-fold validation procedure was applied to test the classifiers with a 4:1 training-testing ratio. Each classifier performance was evaluated based on the misclassification ratio (error).

A part from classification, the histogram of MAV, WL, ZC and SSC from both acquisition systems was computed to evaluate the effect of the different sampling frequencies. For this purpose, histograms were averaged among subjects and normalized for visibility.

Statistics

Non-parametric Friedman's pair test was employed to evaluate the difference between CONV and MYB6, as well as MYB6 and MYB8, using all classifiers. In addition, non-parametric Kruskal-Wallis test was used to compare the two best classifiers within each acquisition system. P-values less than 0.05 were considered significant for both tests.

RESULTS

Table 1 shows the mean percentage classification error and the standard deviation for LDA, SVM, KNN, NB and NN for the three acquisition systems: CONV, MYB6 and MYB8.

Friedman's pair test revealed no significantly different classification performance between CONV and MYB6 (P=0.10). On average, MYB8 showed 1.53 points less than MYB6 in the mean percentage classification error for all classifiers. Nevertheless, this difference did not imply a significant improvement (P=0.16)

NN outperformed all classifiers independently of the acquisition system. When comparing NN with the second best (LDA), NN's performance was significantly better than LDA's in CONV (P=0.02) but not in MYB6 (P=0.17) nor MYB8 (P=0.11).

Table 1: Mean classification error percentage ± standard deviation of CONV, MYB6 and MYB8

	CONV	MYB6	MYB8
LDA	5.82 ± 3.63	9.86 ± 8.05	8.33 ± 6.80
SVM	9.70 ± 6.02	11.01 ± 8.79	9.57 ± 7.63
KNN	8.30 ± 6.00	11.12 ± 8.94	9.61 ± 7.62
NB	12.48 ± 8.51	13.77 ± 9.76	11.95 ± 9.00
NN	1.77 ±1.28	4.64 ± 4.25	3.31 ± 3.37

Figure 2 depicts the computed histograms of the normalized features MAV, WL, ZC and SSC for MYB6 and CONV. Results show a wider and more evenly distributed dynamic range of MAV and WL for MYB6 than for CONV. In contrast, the distribution of ZC and SSC in CONV provides more information than MYB6.

DISCUSSION

Although the application of MYB in advanced myoelectric control is gaining interest in the research community, few studies have been carried out to evaluate its performance. Most of them, are application-oriented [9,10] and do not assess the real capabilities of MYB as an EMG acquisition system, and its potential in pattern recognition. Therefore, the objective of this study was to evaluate the suitability of MYB for hand gesture classification, and compare the effect of its narrow bandwidth with a full bandwidth acquisition systems (CONV).

The first difference between CONV and MYB can be found in the histograms of the features. MYB seems to have a broader range in MAV and WL, which provide information about the amplitude of the EMG signal. However, in the frequency-related features (SSC and ZC), MYB shows a limited dynamic range when compared to CONV. This difference in the frequency information is consistent with MYB's lower sampling frequency. In addition, the non-similar feature distributions may explain the difference in the optimal number of KNN's neighbors: 13 or 15 for MYB (depending on the number of channels), in contrast with the one required for CONV.

The obtained drop in the classification error for LDA and NN using CONV and MYB, is consistent with Li et al. [11] findings. From 1 kHz to 200Hz an approximately 3.5 percentage points drop in the average classification accuracy was found for LDA in able-bodied subjects. However, the classification error of NN and LDA, was found to be lower than in other studies such as Ortiz-Catalan [19]. Using a similar setup, classifying eleven hand motions, sampled at 2 kHz and extracting four time domain features (MAV, ZC, SSC and WL), the obtained classification errors for LDA and NN were 7.9% and 8.8%, respectively.

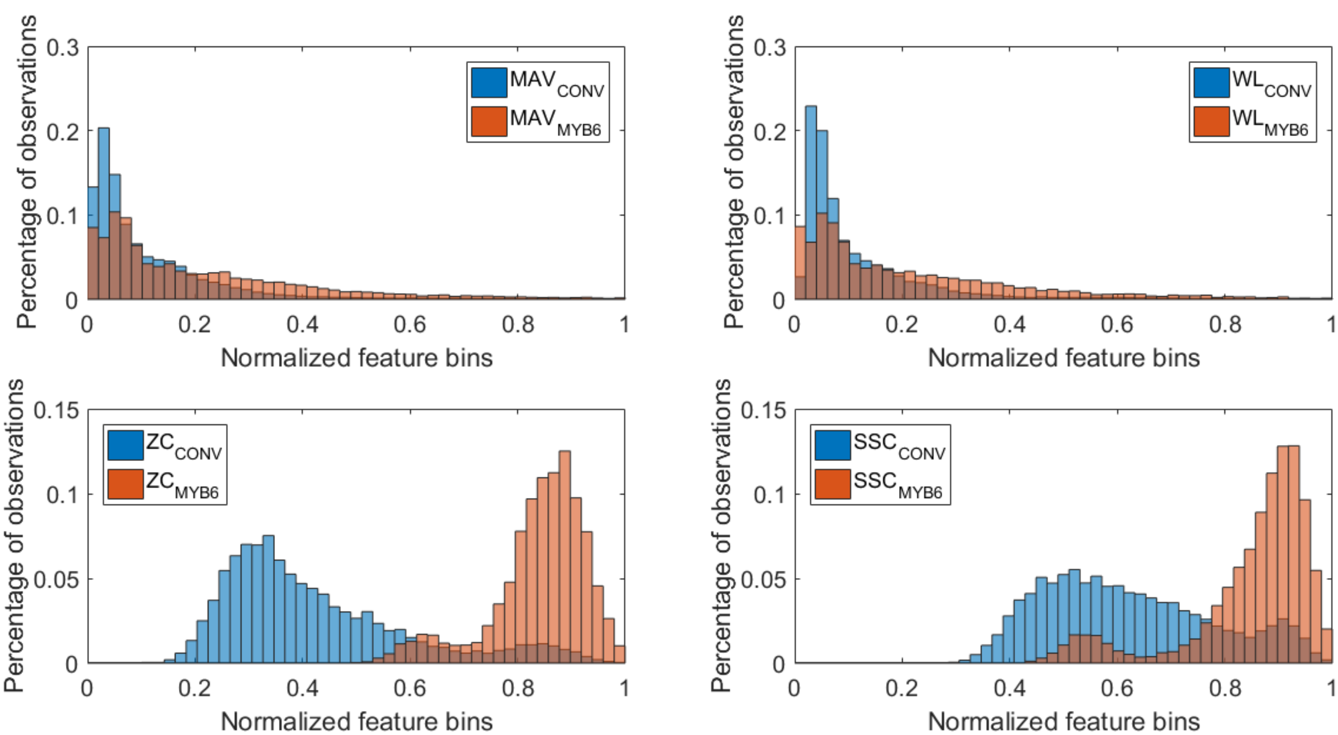

Figure 2: Normalized histograms of the normalized features MAV, WL, ZC and SSC for CONV and MYB6.

Finally, variability in the accuracy due to differences in the number of channels may not be remarkable, as the comparison of MYB6 and MYB8 showed no significant difference, despite the increase in number of channels. This was found to be supported by [20, 21] where increasing the number of channels by two, yielded in little change in accuracy.

CONCLUSION

Offline classification error of MYB and its not significant difference with CONV, demonstrated that MYB is suitable to be used as an EMG acquisition system for pattern recognition applications. Future work should focus on assessing the performance of MYB in an online configuration, and compare it to the CONV standard.

Since the MYB is an intuitive wearable (wireless, with fixed distance between electrodes and no preparation of skin), it could make the acquisition process of EMG data less time consuming and thus, more attractive for upper-limb prosthesis control systems.

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UNDERSTANDING ERRORS IN PATTERN RECOGNITION-BASED MYOELECTRIC CONTROL

Jason W. Robertson, M.Sc.; Kevin B. Englehart, Ph.D.; Erik J. Scheme, Ph.D.

Institute of Biomedical Engineering, University of New Brunswick; Fredericton, NB, Canada

ABSTRACT

Recent advances in pattern recognition-based myoelectric control have allowed the technology to be commercialized after decades of controlled laboratory and clinical usage. Despite this success, challenges remain; one of the most critical of these is susceptibility to unintended movements. These errors often require correction to accomplish the desired task, a frustrating process that can hinder device adoption. Although many studies have examined systemic causes of error, such as limb position, electrode shift, and changes in patterns over time, no study to date has investigated how errors actually occur in real time. A handful of studies have analyzed the training data to establish what, if any, characteristics are predictive of successful control, but with little success. In this work, we examined and characterized the nature of errors as they occurred during a real-time myoelectric control task.

To better understand how errors occur, 24 subjects (50% female, 92% right-handed, age 25.8±3.2 y) were recruited to participate in a myoelectric control task. Subjects elicited eight sample contractions of four movement classes (wrist pronation, wrist supination, chuck grip, and hand open) and a no-movement class, which were used to train a classifier. This classifier was then used to control a cursor through a virtual targeting task, during which the myoelectric signals and the resultant cursor position were recorded. Indices of separability, repeatability, and variability were calculated from the training data, while outcome measures based on Fitts' Law were computed for the usability trials.

A thematic analysis of the real-time errors resulted in identification of three major types of error. The first, overshoot, described when users moved past or through a target without stopping. The second, bounceback, referred to an unintentional activation while the user attempted to stop. The third category encompassed all other active errors that resulted in movement away from the target. Within these error categories, several descriptive metrics were computed, including proportional control values, classifier confidences, and feature space distance metrics.

Although traditional Fitts' law metrics could not be predicted from the training data alone, several of the proposed real-time error characterization metrics could be. Although these measures were calculated during Fitts' Law tests, their results were not found to be correlated with the more traditional Fitts' Law measures. This work suggests that prediction of future performance of pattern recognition-based myoelectric control may be achieved through a better understanding of the nature of errors.

DESIGN OF A POWERED THREE DEGREE-OF-FREEDOM PROSTHETIC WRIST

Neil Bajaj and Aaron Dollar

Yale University

ABSTRACT

Development of upper limb prosthetic devices generally focuses on improving the dexterity or functionality of the terminal device. As a result, currently available wrist prostheses tend to be simplistic devices which cannot replicate the function of the unaffected human wrist. Recent studies have shown that the unaffected wrist contributes to manipulation capability as much as the hand. This implies that a prosthetic wrist which is capable of three degree-of-freedom (DOF) motion may be as beneficial to amputees as complex terminal devices.

In terms of mechanical design of wrist prostheses, the vast majority of devices currently available tend to be passive multi DOF or powered single DOF units. Moreover, the multi DOF units tend to be exceedingly long devices, which may be unsuitable for transradial amputees. Many design innovations borrowed from traditional robotic design and implemented in prosthetic hands could serve to improve the mobility of wrist prostheses or aid in creating compact devices. Achieving full 3 DOF wrist motion in a compact volume is an imperative in wrist design. Thus, herein we present the design of a prosthetic wrist which satisfies this design imperative.

Our design consists of a two DOF mechanism responsible for flexion/extension and radial/ulnar deviation in series with a single DOF pronation unit. Majority of our efforts focus on the development of the two DOF parallel mechanism. We chose a U, 2-PSU architecture for the 2 DOF mechanism and optimized the geometric design parameters of the parallel mechanism in order to maximize a novel metric. Whereas typical parallel mechanism optimization would maximize a dexterity or range of motion based metric, our metric encompasses these as well as resultant size of the mechanism, which is particularly relevant for upper limb prostheses. This results in a compact design with reasonable motion capabilities over the workspace.

INITIAL CLINICAL EVALUATION OF THE MODULAR PROSTHETIC LIMB SYSTEM FOR UPPER EXTREMITY AMPUTEES

[1,3]Kristin Yu, [2]Courtney Moran, [2]Robert Armiger, [2]Matthew Johannes, [3]Paul Pasquina and [1]Michelle Nordstrom

[1]Henry M. Jackson Foundation
[2]Johns Hopkins University Applied Physics Laboratory
[3]Walter Reed National Military Medical Center

BACKGROUND

This study evaluates the performance and usability of the Modular Prosthetic Limb (MPL) in the clinical setting to assess usability and optimize its control and features.

The MPL provides intuitive control schemes to command up to 17 independent joints using real-time embedded software to replicate the functionality of a normal arm and hand, achieved with an array of sEMG sites for intuitive movement via pattern recognition.

METHODS

This is a non-randomized clinical optimization study. Up to 24 upper extremity amputees will be enrolled, with the first 12 to achieve the minimal level of VIE-based pattern recognition control progressing to clinical use. Optimization is based on performance and usability in pattern recognition training and functional improvements with respect to the number of available, controllable degrees of freedom.

This study consists of a virtual training Phase 1 and integrated training-and-use Phase 2. During Phase 1, participants train in a miniature virtual integrated environment to learn pattern recognition.

During Phase 2, prosthetic sockets are integrated with sEMG electrodes to allow users to operate the MPL. Data is obtained during performance of activities of daily living and from standardized, validated self-report and functional assessments administered in clinical sessions.

RESULTS

Seven participants across multiple levels of upper extremity amputation have completed the protocol to date. Participants were able to control a greater number of individual joint and hand motions and increase proficiency with these motions over time. All participants, except one, improved in average motion completion percentages and path efficiencies—normalized by the total number of motions tested—on a Target Achievement Test over time, independent of level or cause of amputation. While speed to complete tasks with the MPL did not approach the speed with a conventional prosthesis, participants utilized a greater number of motions than with their conventional prosthesis and the fidelity of MPL control continued to improve.

CONCLUSION

Participants demonstrated the ability to control a greater number of motions, utilize multiple task-appropriate grasps, and describe a more intuitive control experience than currently available with conventional prostheses. While quantitative functional assessment scores were lower than conventional, MPL use and pattern recognition control improved over time with respect to the number and quality of motions controlled without plateau, indicating further potential function gains.

Disclaimer: The views expressed are those of the authors and do not reflect the official views of the Department of the Army, the Department of Defense, or the U.S. government.

CHANNEL SELECTION OF NEURAL AND ELECTROMYOGRAPHIC SIGNALS FOR DECODING OF MOTOR INTENT

[1]Jacob Nieveen, [2]David Warren, [2]Suzanne Wendelken, [3]Tyler Davis, [2]David Kluger, [2]David Page, [2]Jake George, [4]Christopher Duncan, [5]Douglas Hutchinson, [6]V John Mathews and [2]Gregory Clark

[1]University of Utah, Electrical and Computer Engineering
[2]University of Utah, Bioengineering
[3]University of Utah, Neurosurgery
[4]University of Utah, Physical Medicine & Rehabilitation
[5]University of Utah, Orthopedics
[6]Oregon State University, Electrical Engineering and Computer Science

ABSTRACT

The ability to perform multiple degree of freedom (DOF) proportional control has distinct advantages when applied to next-generation prosthetic hands with individuated finger and wrist control. Regression-based methods provide simultaneously multi-DOF control and can perform untrained hand grasps (Clark et al., MEC17). Such methods use multiple feature channels per DOF but increasing channel count can diminish performance and increase the computational complexity. Ideally, one wants to use the fewest channels that provide the best performance. Here we report a comparison of channel selection methods and recommend a forward stepwise selection method with Gram-Schmidt orthogonalization applied between steps. This approach uses the fewest channels that results in equivalent or no worse performance than other methods investigated, and it is our current standard method for real-time testing.

We used data from one volunteer with transradial amputation to compare the performance of four channel selection methods: channels that correlate with movements (CORR); Gram-Schmidt orthogonalization, forward selection (GS); Least Angle Regression (LARS); and Mutual Information (MI). The subject was chronically implanted with a 32-electrode intramuscular electromyogram (EMG) array in residual forearm muscles (Ripple, LLC) and two Utah Slanted Electrode Arrays (USEAs, Blackrock Microsystems), one in each of the median and ulnar nerves. From these sources, the Mean Absolute Value of 32 single-ended and 496 differential EMG channels and neural firing rate of 192 USEA channels were calculated at 30 Hz (720 total channels). All analyses were performed offline using six online training data sets. From these data, channels were selected and a decoder was trained with subsets of single DOF movements across 6 DOFs and tested with distinct subsets. Performance was quantified by the root mean squared error (RMSE) normalized to each joint's range of motion.

All methods examined performed to similar levels in testing datasets, achieving a minimum mean RMSE of 0.11+/-0.0032 (mean+/-SEM over all DOF and datasets, no significance difference between methods). However, the number of channels necessary to achieve that best mean result differed among channel selection methods with GS requiring the fewest channels (55) and CORR requiring the most (122). MI failed to find an optimal set from 720 channels within reasonable computation time. On an Intel i7, the processing time necessary to select channels ranged from 240 milliseconds (CORR) to 52 minutes (MI, while selecting from 80 EMG channels), with GS taking 24 seconds. From these results, we recommend using the Gram-Schmidt orthogonalization, forward-selection method to choose feature channels.

ELECTRICAL STIMULATION OF THE CERVICAL DORSAL SPINAL CORD AND ROOTLETS FOR SENSORY RESTORATION IN UPPER-LIMB AMPUTEES

Santosh Chandrasekaran, Ameya Nanivadekar, Eric Helm, Michael Boninger, Jennifer Collinger, Robert Gaunt and Lee Fisher

University of Pittsburgh

INTRODUCTION

Numerous studies indicate that sensory feedback could enhance the embodiment, acceptance, and also the ease of use of a prosthetic device. Electrical stimulation of the peripheral and central nervous system is the focus of extensive research as a means to provide sensory feedback. While drawbacks of peripheral nerve stimulation include electrode migration and off-target activation, cortical brain stimulation is an extremely invasive procedure. In contrast, we targeted the dorsal spinal cord and rootlets (DSCR) to provide sensory feedback. This approach affords at least two key benefits. First, the DSCR provide a clear separation between the sensory and motor pathways in the peripheral nervous system. Thus, stimulation at the DSCR will avoid undesired concurrent activation of motor pathways. Second, multiple minimally invasive surgical techniques exist to access the DSCR. In fact, about 50,000 procedures a year are performed in the United States, where spinal cord stimulation (SCS) leads are inserted percutaneously to target the DSCR for alleviating intractable pain. Here, we present observations from human psychophysics experiments performed while stimulating the C5-C8 DSCR in two upper-limb amputees using these FDA-approved SCS leads.

METHODS

All procedures were approved by the University of Pittsburgh Institutional Review Board and the US Army Human Research Protection Office. Two study participants with high-level unilateral upper-limb amputations (>16 years and >5 years post-amputation) were implanted with three percutaneous 16 or 8-contact SCS leads (Boston Scientific) respectively, in the lateral epidural space of the cervical spinal cord. Stimulation was delivered using a customized setup for up to 4 weeks, after which the electrodes were removed. Information regarding the modality, location, and intensity of perceived sensations was provided by the subject using a structured reporting system.

RESULTS

Sensations reported by the subjects included focal percepts localized to the amputated arm, hand, wrist, palm, and fingers. The focality of the sensory percepts could be improved by employing current-steering effects through multi-polar stimulation. Although most of the sensations were reported to be paresthetic in nature, subjects did describe some percepts as touch, pressure and movement of fingers and the arm. The focal locations of the sensations were stable for the entire duration of testing. We found that stimulation frequency had the stronger effect than stimulus amplitude on the intensity of perceived sensations and that it also dictated the perceptual modality of the sensation.

CONCLUSION

With current-steering, DSCR stimulation can generate focal sensory percepts in the missing limb in long-term amputees.

A COMPARISON OF HOME TRIALS WITH MULTIPLE DEVICES AND CONTROLS WITH A SINGLE TH TMR SUBJECT

Todd Kuiken[1,2], Laura Miller[1,2], Levi Hargrove[1,2], Kristi Turner[1], Jon Sensinger[3]

[1] *Shirley Ryan AbilityLab*, [2] *Northwestern University*

[3] *University of New Brunswick*

ABSTRACT

Outcomes can be influenced by both componentry and by the type of control. A case is presented in which one subject completed minimum 6 week home trials with 4 different configurations: commercial arm system with a powered hook with direct control (DC), commercial arm system with a powered hook with pattern recognition (PR) control, a lab developed prosthesis with a powered hand with pattern recognition control and a lab developed prosthesis with a powered hook with pattern recognition control.

INTRODUCTION

Targeted Muscle Reinnervation (TMR) is a surgical technique that increases the number of control signals available as input to a myoelectric prosthesis [1, 2]. This is especially advantageous for higher level amputees who are limited in the number of inputs that might be available. Originally, individuals were limited to using one signal to control one motor movement, DC. With DC, the subject must isolate each individual muscle and the prosthetist then sets individual gains and thresholds for each channel [3].

With pattern recognition, multiple EMG channels can be used as input with the information all considered globally to calculate which "pattern" is being recreated. Since muscle signals do not need to be targeted and isolated, more information can be extracted from the user, potentially increasing the ability to control a multi-degree-of-freedom system [4].

As part of an ongoing study, individuals with a transhumeral amputation as well as TMR were recruited to compare DC to PR using a commercial arm (Boston Digital Arm, with a Motion Control wrist rotator and a single-degree of freedom terminal device, either a hook or hand). Following this first phase subjects were then fit with a system built at the Shirley Ryan AbilityLab, formerly the Rehabilitation Institute of Chicago (RIC), which had a powered elbow, wrist rotator, wrist flexor and hand. However, subjects who had used a hand with the commercial arm phase commented that the hand was not as functional as a hook. Therefore, a powered hook (Motion Control ETD) was fit to the RIC arm and the home trial repeated.

METHODS

The first subject recruited to this study was a 35-year-old gentleman who sustained a R-TH amputation secondary to trauma (military) four years previous. His TMR surgery was performed at the Walter Reed Army Medical Center approximately nine months post-amputation. At the time of recruitment, he was wearing a four-site TMR system with a Dynamic Arm, Wrist rotator and ETD. The project was approved by the Northwestern University Institutional Review Board and written informed consent was obtained from the subject.

As the first subject to take home a pattern recognition TH-TMR system, his participation was not randomized. This choice was made to confirm success of the control system in a home environment for the remaining subjects. This individual had experience with multiple laboratory fittings of pattern recognition systems leading up to the development of a clinically viable configuration.

The subject then completed a home trial with the commercial arm system in four-site direct TMR control. The two TMR sites controlled hand open and hand close. The native biceps and triceps controlled elbow up and elbow down with an elbow flexion impulse signal switching control to his wrist rotator to mirror his home DC system.

Originally, the final phase of the study was the fitting of an RIC developed prosthesis. The design specifications for this device were to be small and light while matching the performance of commercially available components. The device had powered motors at the elbow, wrist rotation, wrist flexion and hand (Figure 1). The hand had a motor in the thumb and one in the fingers. This allowed multiple positions but the two grips used were a tripod, where the thumb would move to a position and stop to oppose the index and middle as they closed, and a power, where the fingers would close and then the thumb close around the fingers.

It was later noted that by this subject and others that had used a hook commented that tasks were much more difficult with a hand than a hook. It was suspected that this might obscure the benefits of an added wrist flexor so the RIC developed prosthesis was modified to allow the use of an ETD and the home trial was repeated (Figure 2).

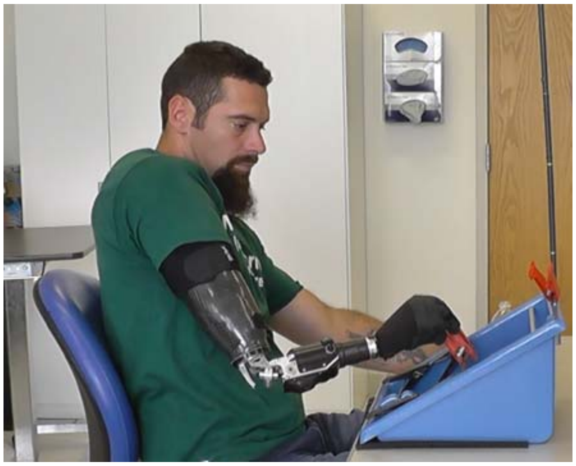

Figure 1: Subject completing the Clothespin Relocation task with the RIC arm/hand system

Figure 2: Subject completing a subtest of the Jebsen using the RIC arm/ETD system.

Outcome measures were collected with all systems pre- and post-home trial. These included the Southampton Hand Assessment Procedure (SHAP), Box and Blocks, Jebsen, Clothespin and, post-home only, the ACMC [5-9]. Each home trial for each of the 4 configurations lasted a minimum of 6 weeks.

RESULTS

All pre-home outcomes have been completed and are presented. Post-home outcomes have been completed for the first 3 systems and are scheduled to be completed for the final system by May 1. ACMC results for the three completed home trials are listed in table 1.

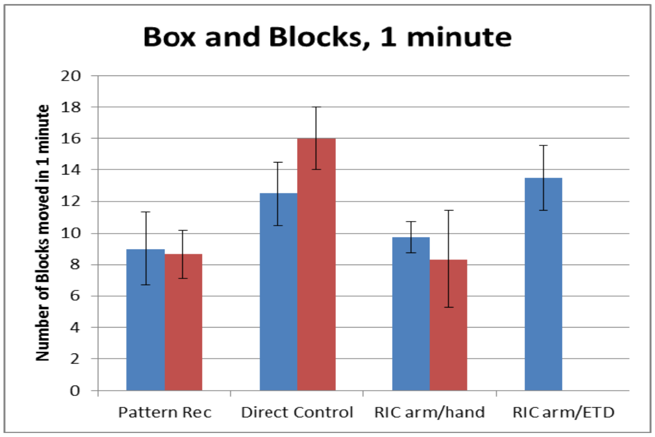

Figure 3: Number of blocks moved in 1 minute for each of the 4 configurations (average/standard deviation of 3 trials per configuration). Pre-home results are on the left and post home results are on the right for each.

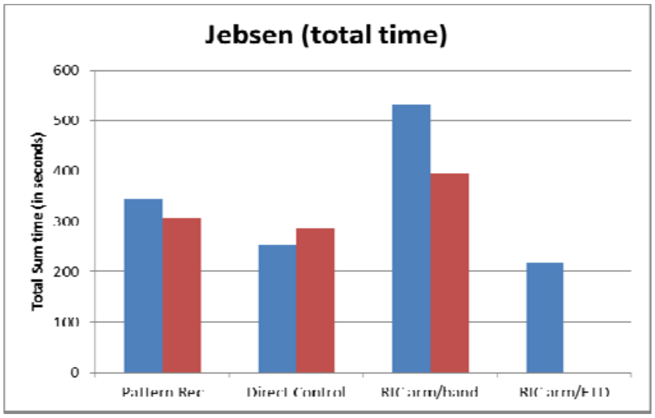

Figure 4: Total sum time of the seven subtasks for each of the 4 configurations. Each subtasks had a max of 120s. Pre-home results are on the left and post home results are on the right for each.

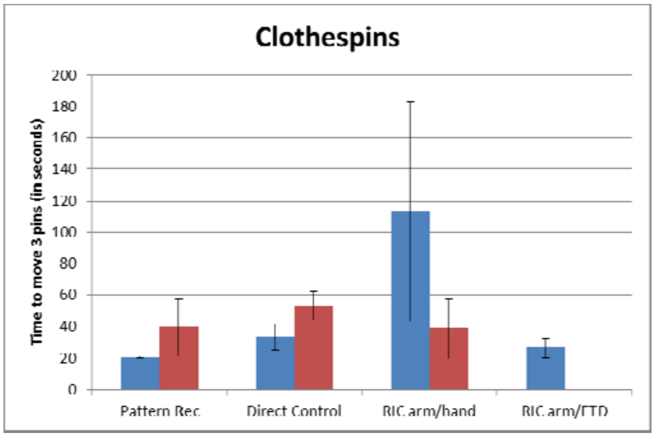

Figure 5: Total time to move 3 clothespins from a low horizontal bar to an upper vertical bar (average/standard deviation of 3 trials per configuration). Pre-home results are on the left and post home results are on the right for each.

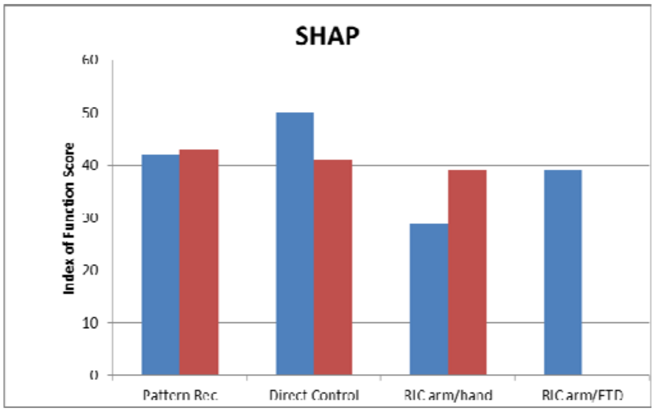

Figure 6: SHAP index of function score. Each subtasks had a max of 120s. Pre-home results are on the left and post home results are on the right for each.

Table 1: ACMC results (post-home trial)

Home trial configuration	***ACMC score***
Commercial Arm with PR	67.6
Commercial Arm with DC	62.0
RIC arm/hand (with PR)	41.3

Subjectively, PR was preferred over DC for the commercial arm. The subject's specific feedback was "I felt I had better control with pat rec with less fatigue. I would also prefer not to have to switch modes" but that both PR and DC were "easy" to make move when he wanted (5/5 for both). During the DC phase he complained of inadvertent movements (hand open/elbow extension with strong signals). With the RIC arm, he gave feedback "the wrist flexor was great" and that "hooks are better" than hands and observationally he used the wrist flexor frequently in both the RIC arm hand and hook systems.

DISCUSSION

The results show that for simpler tasks, where only one degree-of-freedom might be needed at a time, such as box and blocks, there is improved performance with direct control which is less likely to result in inadvertent movement of the wrist rotation since an intentional switching is required to access this additional movement. With tasks where wrist rotation is clearly helpful, such as the clothespin relocation task, there is improvement in function with pattern recognition control where it is easier to access wrist function without switching. This user performed well on most outcomes and, with the more complex tasks, such as the Jebsen and SHAP, performed equally well.

When evaluating the RIC arm/hand, there were notable differences in the clothespin task pre and post home trial. This was partially due to learning to better control the additional degree-of-freedom (wrist flexion) and to accommodate to the hand, but there were updates made to the finger shape during the trial to improve the pinch.

Additional improvements have been made to the device since the RIC arm/hand home trial including updates to the motor controllers to allow for smoother slow control of the wrist rotator and wrist flexor. With these improvements as well as the use of ETD, it is expected that there will be a trend towards improvement.

Final results will be available for presentation at the meeting and it is expected that the subject will be available to demonstrate the function of the full system at the meeting as well.

ACKNOWLEDGEMENTS

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CONTROLLER SELECTION FOR MYOELECTRIC PROSTHETIC HANDS

Eric McClain and Sanford Meek

University of Utah

ABSTRACT

Powered prosthetic hands are traditionally controlled using proportional control, where the motor voltage varies in proportion to the differential EMG signal form antagonistic muscles. This method of control performs adequately in position control as well as grasping durable objects perform poorly where force control is needed, such as grasping brittle objects. We are seeking to improve the ability to control grip force of a hand prosthesis through modifications to the control scheme for both a basic and a responsive prosthetic hand. For the basic hand in this study we used an Ottobock MyoHand with direct motor drive. For the responsive hand internal feedback control was implemented on the MyoHand to compensate for the high effective inertia and friction of the system. The two modifications under consideration are EMG gain scheduling and adaptive time constant low pass EMG filtering. EMG gain scheduling is a simple scheme which consists of changing the EMG signal gain to one of two user selected values based on the activity. One value is for control of an open hand and the other for control while an object is grasped. The adaptive EMG filter varies the time constant of a low pass filter based on the control signal to allow the filtered signal to track fast control signals while filtering out noise when the EMG signal is relatively constant.

Two experiments were performed to compare the variations of myoelectric prosthetic hand controllers. Control of force, control of position and manipulation of a brittle object were evaluated. The manipulation task was performed using a manipulandum that slips at low grasping force and breaks with excessive grasping force. Force and position tracking were evaluated by the ability to track desired values displayed to the subject. Users self-select separate EMG gains for force and position control. Evaluations were performed with both fixed and scheduled gains. The adaptive EMG filter was compared against a fixed time constant low pass filter for each of these conditions. For the basic (Ottobock MyoHand) it was found that the adaptive filter showed no significant improvement but EMG gain scheduling showed a significant increase ($p<0.05$) in performance and user rating of the brittle object manipulation. For the responsive hand (Ottobock MyoHand with internal control) it was found that EMG gain scheduling showed no significant improvement but the adaptive filter showed a significant increase ($p<0.05$) in performance and user rating in force control and brittle object manipulation.

A STUDY INVESTIGATING TARGETED MUSCLE REINNERVATION FOR INDIVIDUALS WITH TRANSRADIAL AMPUTATIONS

Todd Kuiken [1, 2], Kristi Turner [1], Laura Miller [1, 2], Annie Simon [1, 2]

[1] *Shirley Ryan AbilityLab, Chicago, IL, USA*

[2] *Northwestern University, Chicago, IL, USA*

Current strategies do not allow individuals with a transradial amputation to fully benefit from newly available multi-articulating hand prostheses. Targeted Muscle Reinnervation (TMR) surgery, where residual nerves are transferred to target muscle sites, has been successful in providing additional neural control information for higher-level amputees [1]. Additionally, pattern recognition (PR) of residual limb muscle signals has provided advanced control of multifunction prostheses. The objectives of this study were to quantify and compare PR control of a multi-articulating hand before and after TMR surgery in transradial amputees.

Previous myoelectric users with a unilateral transradial amputation were recruited and enrolled at the Shirley Ryan AbilityLab and Walter Reed National Military Medical Center. Subjects were fit with a custom socket with eight bipolar electromyography (EMG) channels, a passive wrist, modified Touch Bionics i-limb revolution hand, and a Coapt Complete Control System. The study was divided into three 8-week home trials: pre-TMR conventional control, pre-TMR PR control, and post-TMR PR control. Subjects participated in the pre-TMR home trials in a randomized order. For the TMR surgery, the median nerve was transferred to the flexor digitorum superficialis muscle and the ulnar nerve to the flexor carpi ulnaris muscle. Subjects participated in the post-TMR home trial at least 6 months post-surgery. Prior to starting each home trial, subjects were trained with an Occupational Therapist. While at home, they complete a daily log of their wear time, usage, and level of control of the device. At the end of each home trial, a variety of outcome measures were scored including the Southampton Hand Assessment Procedure and the Assessment of Capacity for Myoelectric Control.

Currently, three subjects are participating in home trials. The grips were selected with the help of the occupational therapist to include those most functional for a variety of daily activities. With pre-TMR conventional control, all subjects were able to select up to five grips using four triggers. With pre-TMR PR control, the two subjects who have begun home trials have selected four grips (Tripod, key, power, and precision pinch open). While the subjects have had 3-5 grips available, mainly two grips were used. Current use times reported for the home trials averaged 4.5 hours/day for subject one and 2.2 hours/day for subject two. Post-TMR PR home trial results will be discussed as well as any differences seen in the level of prosthesis control and/or performance compared to the pre-TMR PR home trial.

ACKNOWLEDGEMENTS

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PROSTHESIS GRIP FORCE MODULATION USING NEUROMORPHIC TACTILE SENSING

Luke Osborn[1], Harrison Nguyen[1], Rahul Kaliki[1,2], Nitish Thakor[1,3]

[1]*Department of Biomedical Engineering, Johns Hopkins University, Baltimore, MD, USA*
[2]*Infinite Biomedical Technologies, Baltimore, MD, USA*
[3]*Singapore Institute for Neurotechnology, National University of Singapore, Singapore*

ABSTRACT

For many prosthesis users, the lack of tactile feedback in their limbs can make grasping difficult, especially when handling delicate objects. The lack of tactile feedback is not a new problem for prosthesis users, but how tactile information is handled can have a significant impact on the performance of the system. As prosthetic limbs become more advanced, there is an interest in developing systems that are biologically inspired to more closely mimic how the healthy human system operates. In this work, we utilize a leaky integrate and fire neuron model with spike rate adaption for representing tactile information in a prosthetic hand. We investigate the use of the simulated neuron spike rate in an EMG gain modulating function to limit the amount of grip force applied by a prosthetic hand during grasping of a delicate object. We compare this method with the use of the grip force as an input to the EMG gain modulating function as well as to grasping with no tactile feedback. Results show a reduction in the percentage of broken objects during grasping from 27.5% with no feedback to ~14% when using either grip force or neuromorphic spiking feedback. This demonstrates the feasibility of using a biologically relevant representation of tactile information for improving prosthesis functionality in real-time.

INTRODUCTION

The sense of touch offers a multitude of functionality such as exploring intricate objects, performing complex finger movements, or even providing comfort to loved ones. The seemingly unparalleled performance of tactile sensation gives rise to our instinctive behavior to reach out and explore new objects or surroundings with our hands. Our sense of touch helps provide information on texture, shape, weight, and temperature, which we rely on for understanding objects [1]. One problem faced by people with upper limb loss is the lack of tactile information in most commercial prosthetic limbs available today [2]. Although recent developments in myoelectric (EMG) prosthesis control have shown improvements in pattern recognition control strategies [3]–[5], a major component of creating fully functioning upper limb prostheses is tactile feedback. This has led to progress in novel closed-loop tactile feedback control algorithms [6], [7] and sensory feedback via peripheral nerve stimulation [8]–[10].

As technology moves towards more human-like prosthetic arms it is necessary to develop faster, more efficient, and more natural ways of processing tactile information to be used for sensory stimulation. Early work with sensory feedback of tactile information used force sensor information to drive peripheral nerve stimulation where increased grip force translated to increased stimulation frequency, which was used for object discrimination [8] and grip force modulation [9]. More recently, a neuromorphic stimulation model was implemented using signals from a tactile sensing prosthetic finger for texture discrimination [10]. There is a trend towards developing neuromorphic devices and models to mimic the natural behavior of biological systems to improve efficiency and performance over traditional methods. Recent examples include the vestibular system [11], cortical neurons [12], and touch [13]. For tactile feedback in upper limb prostheses a neuromorphic approach includes modeling of the slowly adapting (SA) and rapidly adapting (RA) mechanoreceptors found in our skin. The goal being that this approach will offer more efficient transmission of relevant tactile information, similar to a healthy peripheral nervous system, to the prosthesis controller as well as for driving nerve stimulation for sensory feedback. Previous work using models to simulate tactile afferent patterns have investigated implementation of the models with little emphasis on real-time functionality [14], [15]. Here we investigate the ability of a prosthesis controller to functionally interpret a neuromorphic model of tactile information using a leaky integrate-and-fire (LIF) neuron with spike rate adaption to estimate grip force and prevent breaking a delicate object during a prosthesis grasping task.

MODEL & METHODS

One particular model that is commonly used to simulate the behavior of SA and RA mechanoreceptors is the leaky integrate and fire (LIF) neuron model [14]–[16]. For this work, we implemented an LIF neuron model with spike rate adaption, which introduces a hyperpolarizing current that makes the neuron less likely to fire once it has previously fired. This adapted model is used to create the neuromorphic response and represent a more realistic neuron spiking behavior. The model can be written as

$$\tau_m \frac{dv}{dt} = v_r - v(t) + RI(t) - g(t)(v(t) - E_k) \quad (1)$$

where $v(t)$ represents the membrane potential at time t, and τ_m is the membrane time constant. R is the membrane resistance. This is a simple RC circuit where the leakage is due to the resistor and the integration of $I(t)$ is from the capacitor in parallel. When the membrane potential reaches a spiking threshold, v_{th}, it is reset instantaneously to a lower value, v_r. The refractory conductance of the neuron is given by $g(t)$ and E_k is the reversal potential for the spike rate adaption. The change of the conductance is given as

$$\tau_g \frac{dg}{dt} = -g(t) \quad (2)$$

where τ_g is the conductance refractory period. The conductance is incremented by Δg after each spike. A more detailed and complete discussion of this model and its extensions can be found in [16].

To create a neuromorphic tactile feedback system, we use the output of force sensors as the input stimulus, $I(t)$, to the model. The model is tuned so that the maximum firing rate is 100 Hz, which occurs when the grip force of the prosthesis is 20 N. This model represents a SA type neuron due to its sustained response to a given input. The neuromorphic tactile feedback method presented here differs from our previous work in that it uses more realistic, continuous neuron model dynamics to simulation spiking behavior. The neuron firing rate of the mechanoreceptor model is used to determine grip force, which is then used to prevent accidental damage to delicate objects during grasping. Our previous work utilized event-based spikes to trigger the onset, offset, and changes in force but used the raw sensor signal for determining grip force [6].

The sensors are placed on the thumb, index, and middle fingertips of a bebionic3 prosthetic hand (Steeper, Leeds, UK) (Fig. 1a). The sensors are force sensitive resistors made up of stretchable textiles. These piezoresistive sensors (Fig. 1b) have been previously developed and used for measuring grip force on a prosthetic hand [6], [17]. Each fingertip cuff has 3 sensing elements. A custom control board developed by Infinite Biomedical Technologies (Baltimore, USA) is used to interface with the prosthesis and read in the fingertip force sensor signals. The grip force is found by summing the output of the sensing elements. Electromyography (EMG) electrodes (Infinite Biomedical Technologies, Baltimore, USA) are used to record motor neuron activity in the forearm from the prosthesis user to control the hand. The neuromorphic model is implemented using MATLAB

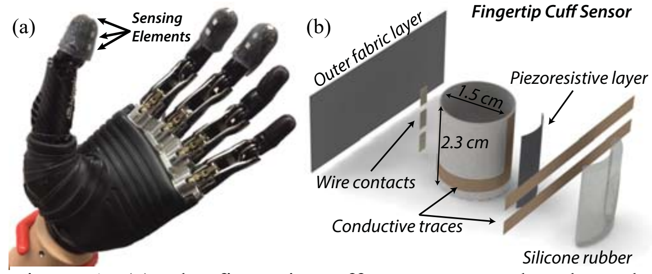

Figure 1: (a) The fingertip cuff sensors are placed on the thumb, index, and middle fingers of a bebionic3 prosthesis. Each sensor cuff contains three sensing elements. (b) The cuff is made up of conductive and piezoresistive textiles as well as silicone rubber.

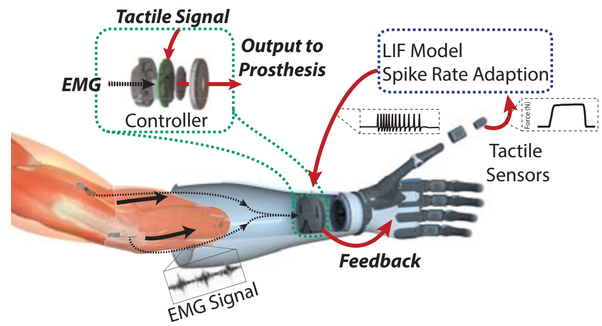

Figure 2: The prosthesis grip force serves as the input to the neuromorphic model. The prosthesis controller processes both EMG and tactile signals, which allows for efficient modulation of the information being sent to the prosthesis as feedback. In this work, the feedback to the prosthesis is a modulated EMG gain that is dependent on the spike rate of the neuromorphic model.

(MathWorks, Natick, USA). The sensor signals are relayed via Bluetooth communication to MATLAB from the prosthesis controller. The sensor signals are sampled and sent to MATLAB at 200 Hz while the EMG electrodes are sampled at 1 kHz. The system diagram is shown in Fig. 2.

An EMG gain modulating function that uses tactile information is implemented on the prosthesis controller to limit the amount of grip force applied during grasping. This exponential decaying function, the Compliant Grasping algorithm, was presented and described in detail in [6]. We adapted the algorithm for this work to limit the maximum grip force to 10 N before forcing the EMG signal to zero. EMG modulation was only applied to the electrode signal that closed the prosthesis. Two algorithm conditions were investigated in this work. The first uses the measured grip force as the input to the EMG modulating function, which is similar to the approach in [6]. The second method uses the output of the neuromorphic model and the neuron firing rate as the input to the EMG modulating function. The goal here is to investigate the ability of a prosthesis to utilize neuromorphic input to successfully modulate a user's EMG signal to improve grasping of delicate objects.

EXPERIMENTS & RESULTS

To evaluate the neuromorphic tactile feedback system the prosthetic hand was mounted on a stand and controlled by the user's forearm EMG signals. Three male subjects participated in this experiment, a bi-lateral upper limb amputee and two able-bodied individuals. The participants controlled the prosthesis to grab, hold, and release a delicate object presented by the experimenter. The experiment was approved by the Johns Hopkins Medicine Institutional Review Board. The goal was to not break the object, a cracker ($m = 1.80 \pm 0.11$ g, force to break > 8 N), during grasping. Each user was allowed to practice with the system for up to 10 minutes before starting the experiment. Three different conditions were tested: 1) no tactile feedback, 2) grip force (GF) tactile feedback and 3) neuromorphic spike rate (SR) tactile feedback. Each trial consisted of 10 presentations of the delicate object, and the number of broken objects was recorded. Up to 10 trials of each condition were performed in a random order. Results from all participants are similar and were combined to provide a larger data set. As described in the previous section, both of the tactile feedback conditions reduced the amplitude of the EMG signal to close the prosthesis, effectively limiting the hand's ability to exert a large grip force, similar to what has been described in [6] and [18].

The neuromorphic response to the tactile signal during grasping is shown in Fig. 3. This figure shows a representative grasp, hold, and release for a single trial from the experiment. The spike rate of the neuron is found using a 60 ms sliding window and is used in the EMG gain modulation algorithm for limiting the amount of grip force applied by the prosthesis. The results from the prosthesis grasping task are shown in Fig. 4. The number of broken objects are recorded and the average percentage of broken objects for each testing condition are shown in Fig. 4. With no tactile feedback, 27.5% of the objects broke during grasping. Using the total grip force as an input to the EMG gain modulation function, 14% of the grasped objects broke whereas 14.5% of the objects broke when using the neuromorphic spiking behavior from the neuron model as the input for EMG gain modulation. The error bars in Fig. 4 represent the standard error of the mean.

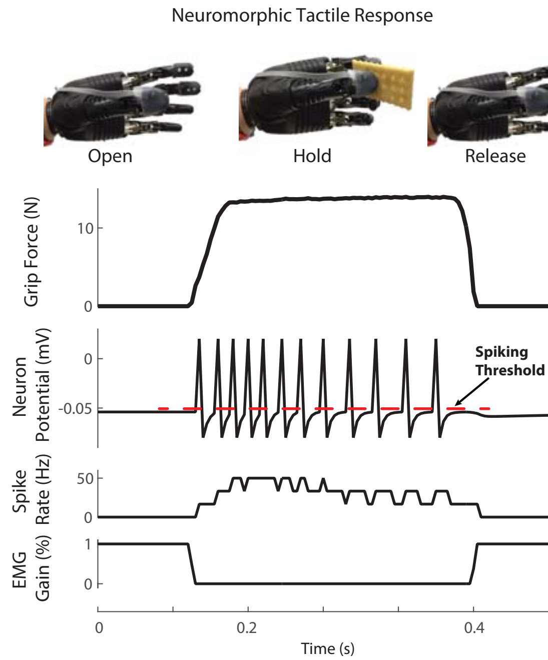

Figure 3: The grip force and neuromorphic spiking response during an actual prosthesis grasping task are shown by the top two curves, respectively. The neuron spike rate and the modulate EMG gain are shown by the bottom two curves, respectively. This data is taken from a single grasping task and is representative of the data set.

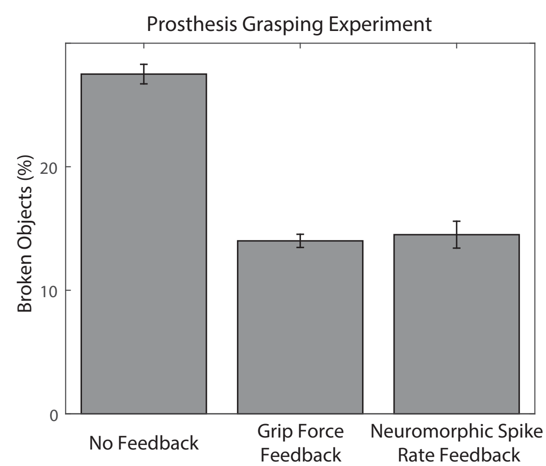

Figure 4: Results from the prosthesis grasping task show improvements while using tactile feedback. The use of grip force or neuromorphic spike rate show improvements over the case of no tactile feedback. The neuromorphic approach shows similar improvements as the more traditional method of using grip force as a feedback input.

DISCUSSION & CONCLUSION

The LIF neuron model with spike rate adaption produces biologically relevant signals with realistic dynamics as shown by Fig. 3. Using neuromorphic feedback achieves similar results as using standard grip force as a feedback mechanism (Fig. 4), but the benefit is in the to process a digital representation of touch. This neuromorphic representation of tactile information is valuable because it allows for transmission of larger amounts of data in a more efficient manner, similar to behavior in biology [1]. The spike rate adaption component of the traditional LIF model provides more realistic neuron behavior by adjusting the neuron conductance with sustained stimulation. This adaption is seen in the prosthesis implementation by the decreasing firing rate during the sustained grip in Fig. 3.

The neuromorphic approach to processing tactile information shows improved performance over no tactile feedback. With no form of tactile feedback, the prosthesis grasping task resulted in 27.5% of the objects being broken during the experiment. Including grip force information as part of an EMG modulating strategy drastically improves this number by reducing it to 14%, which is similar to results seen in [6] and [18]. The average percentage of broken objects is 14.5% while using only the firing rate of the LIF neuron with spike rate adaption for modulating the EMG gain. This is an interesting finding in that it demonstrates the ability of the prosthesis hardware to process the spiking response and transform it into EMG gain modulation. Additional user testing under more scenarios is necessary to better understand the system's performance. The results presented here have major implications for future prosthetic limbs incorporating sensory feedback to the user. Providing realistic neuron activity to the prosthesis will help streamline the information flow from sensors back into the nervous system of the user.

The goal of this work is to demonstrate the feasibility of a neuromorphic tactile feedback system for use in a prosthetic arm. The results from the prosthesis grasping task suggest the ability to use a purely neuromorphic representation of a tactile signal for improving grasping of delicate objects. This is one of the first implementations of a neuron model to represent tactile information for real-time processing by a prosthetic limb. The highlight of this work is the use of a neuromorphic tactile feedback system based on a LIF neuron model with spike rate adaption for real-time functional improvements in a prosthesis. This will play an important role for future prosthetic technology as limbs become more sophisticated and attempt to mimic the human body in both utility and performance.

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COMPARISON OF FUNCTIONALITY AND COMPENSATION WITH AND WITHOUT POWERED PARTIAL HAND MULTI-ARTICULATING PROSTHESES

Andrea B Wanamaker[1], MS, Lynsay R Whelan[2], MS, OTR/L, Jeremy Farley[2], CPO/L,
and Ajit MW Chaudhari[1], PhD, FACSM
[1]The Ohio State University, [2]Touch Bionics by Össur, Columbus, OH, USA

ABSTRACT

This study aimed to evaluate differences between Southampton Hand Assessment Procedure (SHAP) outcome measure scores and kinematic movements during functional tasks for individuals with partial hand limb loss with and without a myoelectric prosthesis. The results presented here indicate that an externally powered hand prosthesis restores function to individuals with partial-hand limb loss, as demonstrated by improved SHAP scores and changes in upper limb kinematics. The kinematic analysis of three functional tasks resulted in the prosthesis condition having decreased upper limb joint range of motion (ROM) compared to the non-prosthesis condition.

INTRODUCTION

The function of the human hand is essential for communication, independence in self-care, and maintenance of a good quality of life. According to the American Medical Association guidelines, amputation of the 4 fingers and the thumb, leaving the palm intact, can result in an impairment rating of 54% of the whole person, while amputation of the entire leg at the level of the hip results in an impairment rating of only 40% [1]. Of the 20,000 new cases of upper limb loss each year, 90% occur at or distal to the level of the wrist [2]. In those individuals with upper limb loss, 50% report overuse problems in their unaffected limbs [3]. Risk factors associated with overuse injury include repetitive motions, awkward joint posture, prolonged unnatural posture, and high gripping force [4]. Use of an upper limb prosthesis may reduce the compensatory mechanisms, particularly awkward and prolonged joint postures, which individuals with upper limb loss use in their activities of daily living.

The Southampton Hand Assessment Procedure (SHAP) is a standardized assessment that measures hand function utilizing both abstract objects and activities of daily living [5]; and has been used in several studies of prosthesis performance. The SHAP results in six general assessment scores out of 100 for the six specific grip styles (spherical, tripod, power, lateral, tip, and extension) and one overall function score (Index).

The objective of this research was to examine differences in SHAP scores and compensatory motions at the wrist, elbow, and shoulder used by individuals with partial hand amputations in completing food preparation tasks both with and without their prosthesis. We hypothesized that there would be significant differences in kinematics between conditions, with the prosthesis condition resulting in reduced joint motion, and joint motion more similar to healthy controls than without the prosthesis. For the SHAP analysis, we hypothesized that the individuals with partial hand amputation would score higher, indicating better function, during the prosthesis condition compared to non-prosthesis.

METHODS

Subjects

Participants were recruited from the Touch Bionics' facility during a week-long training program they attended with their treating prosthetist from facilities across the United States. With the permission of the treating prosthetist, the client was given information on the study and completed testing after providing IRB approved informed consent. Potential participants in the limb loss group had to have an acquired partial-hand amputation (4-digit loss with intact thumb or 5-digit loss), be over 18 years of age, have a minimum of 10 hours of occupational therapy, have good skin integrity, strength, and control of the residual limb (minimum 40º wrist range of flexion/extension, minimum 60º forearm range of supination/pronation, and minimum lateral thumb pinch strength of 10 pounds, if applicable).

Healthy two-handed participants were recruited from the local community. Healthy controls had to be over the age of 18 years and were disqualified if they had previously had surgery on their upper extremities or experienced a severe upper extremity injury, such as a torn muscle, ligament or tendon, or a displaced fracture.

Data Collection

Data collection involved 3D motion capture (Vicon) while participants performed various tasks of daily living focusing on upper limb movement as well as the SHAP which requires the participants to complete the tasks as quickly and accurately as possible while self-timing [5]. The prosthesis users performed the testing protocol twice: once with their prosthesis and once without. The two-handed participants also performed the protocol twice: once with their dominant hand and once with their non-dominant. Condition order was randomized with a fifteen-minute break between conditions to minimize possible learning and fatigue effects. Two trials were collected for each task within each condition, with the best used for analysis

Data presented here reflect the SHAP outcomes and kinematics from the following three food preparation tasks: slicing a tomato, peeling a cucumber, and cutting a piece of meat. To complete the tomato task, participants used a paring knife with a cutting board to slice a tomato four times. The tomato was sliced in half with the flat side on the cutting board to minimize stability issues. To complete the cucumber task, participants use an ergonomically designed vegetable peeler and were instructed to make approximately 4-7 smooth peeling strokes on the cucumber. To compete the meat cutting tasks, participants were given a dinner plate with a ¼ inch piece of cooked breakfast ham. Using a butter knife and dinner fork, they were instructed to cut four bite size pieces. During the prosthesis condition, the users were instructed to use the knives and peeler with their prosthesis and use their contralateral hand solely for support (e.g., situating knife in prosthesis, holding cucumber). For the non-prosthesis condition, the 4-digit users were instructed to try to use the tools with their residual palm and thumb, while the 5-digit users performed the tasks using their contralateral side with the residual limb for support. Healthy controls used the tools in both hands dependent upon condition.

Data Analysis

Upper limb kinematics were calculated using custom Visual3D code. Marker data were filtered using a fourth order low-pass Butterworth filter at 6 Hz. Peak and ROM joint angles for the shoulder, elbow, and wrist were found for each participant for each condition. SHAP function scores were calculated from the timing results, where a score less than the normal 95-100 indicates a functional deficit [6]. Kinematic analysis was limited to when the participant and tools were in contact with the food. Only the movement strokes used to slice the tomato, peel the cucumber, and cut the meat were examined. Statistical comparison between conditions was completed using a mixed-effect model with a between-subjects (individuals with limb loss vs. two-handed healthy) and a within-subject (prosthesis vs no prosthesis; dominant vs non-dominant) design with a significance level of 0.05.

RESULTS

Twelve male individuals participated in the study with the following demographic data (Table 1):

Table 1: Demographic data
* indicates significant difference between groups ($p<0.05$)

	Individuals with Limb Loss		**Two-handed Controls**
Number of participants	4 digit n=3	5 digit n=3	n=6
Age	27.7±8.1 yrs	43.7±18.6 yrs *	25.5±3.9 yrs *
Height	177.8±9.2 cm	173.5±6.4 cm	182.9±5.8 cm
Weight	76.2±.4 kg	97.2±25.3 kg	83.0±13.3kg
Dominant-Side Amputation	3	1	No amputation, all RHD

SHAP Results

As a group, the prosthetic users demonstrated no significant differences in the SHAP scores. The 4-digit subgroup also demonstrated no significant differences (Table 2). The 5-digit subgroup had significantly different scores in four of the six categories, as well as the overall Index score, and trended towards differences in the last two ($p<0.10$) (Table 2). The healthy cohort demonstrated significantly different scores between dominant and non-dominant conditions, and compared to those with amputations, the two-handed participants scored significantly higher on all scores, regardless of condition for either group.

Table 2: SHAP scores for 4-digit and 5-digit groups
Results presented as group mean (SD)
* indicates $p<0.05$ between conditions

	4-Digit (n=3)		***5-Digit (n=3)***	
Category	**Prosthesis**	**Non-Prosthesis**	**Prosthesis**	**Non-Prosthesis**
Spherical	87.0 (5.3)	91.7 (1.5)	**77.7 (2.3) ***	**37.0 (2.6) ***
Tripod	80.7 (9.3)	82.3 (8.5)	53.3 (18.0)	14.7 (10.7)
Power	90.7 (8.7)	87.7 (6.5)	**67.7 (12.7) ***	**14.7 (4.0) ***
Lateral	91.0 (7.8)	93.3 (2.1)	**70.7 (18.1) ***	**11.7 (0.6) ***
Tip	77.0 (10.4)	60.7 (28.1)	46.0 (26.5)	10.0 (6.1)
Extension	94.0 (4.4)	96.7 (0.6)	**77.0 (10.4) ***	**31.7 (19.4) ***
Index	90.7 (5.9)	89.7 (5.7)	**71.7 (13.7) ***	**25.3 (5.9) ***

Kinematic Results

Analysis of shoulder, elbow, and wrist range of motion (ROM) resulted in various significant differences within subjects (prosthesis vs non-prosthesis; dominant vs non-dominant) and between subjects (limb loss vs healthy) for all three food preparation tasks (Table 3). One 4-digit participant was unable to complete the tomato slicing and meat cutting tasks using his residual limb and instead had to use the intact contralateral side. Thus, there were four prosthetic users who were compared to their intact hand for the tomato slicing and meat cutting.

Table 3: Kinematic results for tomato slicing, cucumber peeling, and meat cutting. Presented as group mean (SD). Units are in degrees. The following key is used to indicate significant findings ($p<0.05$): ^{a} between prosthesis conditions, ^{h} between healthy conditions, * between dominant and prosthetic user conditions, and ^ between non-dominant and prosthetic user conditions.

Tomato Slicing		Prosthesis	Non-Prosthesis	Dominant	Non-Dominant
Shoulder ROM	Flexion/Extension	20.6 (8.8)	15.4 (4.7)	17.0 (3.5)	19.4 (4.9)
	Adduction/Abduction	13.7 (8.7)	15.7 (3.9)	11.2 (3.1)	12.3 (3.4)
	Int/Ext Rotation	22.0 (8.7) *^	27.2 (10.7) *^	12.7 (1.9) *	12.2 (2.5) ^
Elbow ROM	Flexion/Extension	24.9 (9.1)	41.3 (20.5) *^	20.8 (5.1) *	21.9 (4.5) ^
	Adduction/Abduction	9.8 (5.8) ^{a}	14.1 (6.8) ^{a}*^	7.7 (2.2) *	7.0 (2.1) ^
	Pronation/Supination	35.1 (27.1) *^	31.6 (5.0) *^	10.1 (1.9) *	9.5 (2.7) ^
Wrist ROM	Flexion/Extension	13.2 (7.2)	16.7 (4.4)	14.0 (3.7)	16.8 (3.3)
	Adduction/Abduction	6.0 (7.2)	8.0 (3.6)	4.5 (1.5) ^{h}	7.1 (2.4) ^{h}
	Hand Rotation	2.4 (0.9) ^{a}^	5.8 (1.2) ^{a}*	2.9 (1.1) ^{h}*	5.4 (1.3) ^{h}^

Cucumber Peeling		Prosthesis	Non-Prosthesis	Dominant	Non-Dominant
Shoulder ROM	Flexion/Extension	13.7 (4.1) ^{a}	18.9 (2.7) ^{a}	16.7 (4.0)	17.5 (4.3)
	Adduction/Abduction	10.5 (4.3)	13.9 (2.7) *^	9.8 (2.7) *	10.1 (2.6) ^
	Int/Ext Rotation	23.8 (6.0)	31.7 (7.8) *^	19.2 (5.1) *	19.7 (8.1) ^
Elbow ROM	Flexion/Extension	27.2 (5.8) ^{a}	39.5 (7.7) ^{a}*^	26.7 (6.1) *	28.9 (3.0) ^
	Adduction/Abduction	15.5 (14.2)	22.2 (15.6) ^	8.4 (1.1)	7.0 (1.5) ^
	Pronation/Supination	33.0 (6.8)	49.6 (25.1) *	22.1 (11.1) *	22.5 (19.1)
Wrist ROM	Flexion/Extension	11.4 (5.9) ^	21.4 (11.3)	15.3 (5.6)	19.5 (6.0) ^
	Adduction/Abduction	5.5 (1.2) ^{a}^	11.3 (3.0) ^{a}	9.0 (4.1) ^{h}	9.3 (3.9) ^{h}^
	Hand Rotation	2.5 (1.3) ^{a}^	5.2 (3.1) ^{a}	2.6 (1.3) ^{h}	9.3 (5.2) ^{h}^

Meat Cutting		Prosthesis	Non-Prosthesis	Dominant	Non-Dominant
Shoulder ROM	Flexion/Extension	29.9 (21.9)	17.4 (3.9)	15.6 (4.0)	21.0 (5.1)
	Adduction/Abduction	17.1 (7.5)	14.3 (5.1)	14.5 (2.8)	14.6 (3.0)
	Int/Ext Rotation	33.8 (22.6)	29.5 (13.7) *	16.3 (2.4) ^{h}*	23.5 (7.0) ^{h}
Elbow ROM	Flexion/Extension	21.8 (10.7)	29.6 (10.7) *	16.8 (1.9) ^{h}*	21.2 (2.5) ^{h}
	Adduction/Abduction	7.4 (2.8) ^{a}	18.4 (13.1) ^{a}*^	6.2 (1.2) *	6.7 (1.2) ^
	Pronation/Supination	41.3 (18.9) *^	44.3 (13.7) *^	14.3 (4.5) *	17.6 (6.8) ^
Wrist ROM	Flexion/Extension	14.0 (10.5) ^{a}	26.7 (10.1) ^{a}*^	12.6 (2.8) ^{h}*	15.7 (2.8) ^{h}^
	Adduction/Abduction	8.2 (3.5) ^{a}	11.6 (2.2) ^{a}*	7.5 (2.6) ^{h}*	9.4 (3.4) ^{h}
	Hand Rotation	3.9 (3.1) ^	7.7 (2.9) *	3.1 (1.3) ^{h}*	8.4 (2.1) ^{h}^

DISCUSSION

The results presented here indicate that use of an externally powered hand prosthesis restores function for individuals with partial hand limb loss. The SHAP outcomes show that the prosthesis improves functional capabilities, particularly for the individuals with 5-digit loss. It should be noted that five of the six individuals with partial hand limb loss demonstrated improved scores with the prosthesis. Only one user with 4-digit loss scored higher without the prosthesis. This individual had exceptional thumb dexterity and his scores without the prosthesis actually fell within the normal range of the SHAP with an intact hand. While the individuals with amputations had improved function with the prosthesis, they still demonstrated significantly lower scores than the healthy cohort. The prosthesis users must consciously open, close, and position the fingers throughout the task, whereas the healthy controls do not have this delay in movement. Thus, while the scores are improved with the prosthesis, the inherent time delay may be why they were not comparable to the healthy cohort results.

The kinematic analysis of various food preparation tasks gives insight to movement strategies with and without the prosthesis. The prosthetic user group had decreased hand rotation, decreased elbow adduction/abduction, and, while not significantly different, had increased shoulder flexion/extension ($p=0.16$) while slicing a tomato. The prosthesis design included a laminated outer frame and silicone inner socket. The trim lines of the socket typically were at or slightly proximal to the wrist, which may have limited wrist motion to some extent. Decreased elbow motion indicates a stiffening of the joint, possibly to improve stability of the knife and wrist during the task. The same strategy of stiffening the elbow and wrist for stability is also seen in the cucumber task, with decreased elbow flexion/extension, wrist adduction/abduction (ulnar/radial deviation), and hand rotation. In the cucumber task, the shoulder also demonstrated reduced flexion/extension. This may be due to how the individual completed the task with their prosthesis, as they tended to use very short strokes to peel the cucumber. Finally, the meat cutting task also indicated reduced wrist and elbow motion compared to the non-prosthesis condition.

Comparison of healthy dominant to healthy non-dominant resulted in multiple differences, particularly with wrist motion for all three tasks. The motion with the dominant wrist appeared more stable and smoother, while the non-dominant side demonstrated a possible lack of control and instability during the cutting tasks, as the non-dominant wrist had increased motion. Comparing the individuals with amputations to the healthy group, there were a high number of differences for each task. The use of the prosthesis did result in motion more similar to healthy dominant than the non-prosthesis condition. The prosthesis condition had seven significant differences with the non-dominant condition across the three tasks. However, five of the seven variables were not statistically different from the dominant condition, indicating that the prosthesis performed more similar to dominant motion than the non-dominant motion. Finally, the non-prosthesis condition was significantly different from both dominant and non-dominant conditions across all three food preparation tasks and had the highest ROMs of all four conditions.

CONCLUSION

The results presented here indicate that an externally powered hand prosthesis restores function to individuals with partial-hand loss, as demonstrated by improved SHAP scores and changes in upper limb kinematics. The kinematic analysis of three functional food preparation tasks resulted in the prosthesis condition having lower joint ROMs compared to the non-prosthesis condition. Comparing to healthy dominant and non-dominant movement, the prosthesis condition was more similar to the dominant condition and the non-prosthesis condition had the highest joint ROMs of all four conditions.

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PATIENT-SPECIFIC CONSIDERATIONS IN IMPLEMENTING ARTIFICIAL SENSORY LOCATIONS

[1]Ivana Cuberovic, [1]Emily Graczyk, [1]Matthew Schiefer, [2]Robert Anderson and [1]Dustin Tyler

[1]Case Western Reserve University
[2]Louis Stokes Cleveland Dept. of Veterans Affairs Medical Center

INTRODUCTION

Sensory restoration is critical for natural prosthesis control. Electrical stimulation of nerves through cuff electrodes restores sensation across the hand. Preliminary studies show that multi-contact stimulation can refine, shift, and create new sensory locations beyond those achievable using single-contact stimulation. Manually tuning these locations is time intensive. Implementing optimized sensations in a clinically viable period requires use of computational models that depend on patient-specific anatomy and cognitive state.

METHODS

To date, four trans-radial amputees have had nerve cuffs implanted on the median, radial, and ulnar nerves. Nerve anatomy and somatotopy was determined using intraoperative ultrasound. After a post-surgical recovery period, subjects underwent a limited mapping study in which they received single-contact stimulation and reported the location of the evoked sensation.

Using the fascicular geometry and initial mapping data, patient-specific models were developed to predict multi-contact stimulation parameters needed to evoke sensation in targeted locations. Finally, subjects underwent a second mapping protocol to map the new and refined sensory locations predicted by these models.

RESULTS

Sensory locations were dependent on cognitive state. Subjects perceived artificial sensations independently of their pre-existing phantom sensation; artificial sensation did not replace the phantom sensation. Thus, we found that the subjects' understanding of their phantom without stimulation affected their perception of the stimulation-evoked sensations. Subjects with poor visualization of their phantom were initially not able to localize sensation on the hand. However, after undergoing visualization training and mirror box therapy aimed at improving subjects' perception of the phantom independent of stimulation, evoked sensations became localized.

Once a clear image of the phantom was established, single contact stimulation elicited unique percepts across the phantom hand for all subjects. As expected, the activated axon population and ensuing sensory location were driven by the fascicular geometry. For example, subjects with more proximal implants reported more diffuse sensations across the hand. Patient-specific models captured these dependencies and enabled us to efficiently determine novel stimulation paradigms in order to evoke sensation in new, functionally relevant locations.

CONCLUSIONS

Providing sensory percepts in functionally relevant locations is dependent on of the patient's nerve anatomy, implant location, and the patient's relationship with their phantom. Patients need a clear understanding of their phantom in order to be able to perceive localized sensations. These locations can be further refined or expanded using computational model derived stimulation parameters.

DIFFERENCES IN INTRAMUSCULAR EMG ACTIVITY IN ABLE-BODIED SUBJECTS AND TRANSRADIAL AMPUTEES DURING STRUCTURED HAND MOVEMENTS

Misagh Mansouri, Carl Beringer, David Weir, Tyler Simpson, Michael Munin, Michael Boninger, Lee Fisher, Jennifer Collinger and Robert Gaunt

University of Pittsburgh

INTRODUCTION

Commercial myoelectric prostheses have limited capabilities to simultaneously control multiple degrees of freedom. These prostheses typically rely on signals recorded from surface EMGs placed on the residual limb, which are not the full set of extrinsic hand muscles required to actuate individual fingers. In addition, standard control approaches usually use pattern recognition or map muscle activity to specific prosthesis movements while largely ignoring underlying biomechanics. Understanding the coordinated activity of extrinsic hand muscles and how their activity results in individual joint movements across a wide range of hand configurations is an essential step towards improving the dexterity of prosthesis control. Here we use dimensionality reduction and clustering techniques to investigate these relationships in able-bodied subjects and an amputee.

METHODS

All procedures were approved by the University of Pittsburgh Institutional Review Board and the US Army Human Research Protection Office. Nine able-bodied subjects and one transradial amputee were recruited for this study. We recorded intramuscular EMG (iEMG) from 16 extrinsic hand muscles targeted using ultrasound. Subjects were instructed to attempt 45 movements that included individual finger and wrist movements in different wrist postures (flexed, extended, pronated, supinated and neutral). iEMG signals were recorded at 30 kHz, high-pass filtered at 20 Hz, rectified, and then low-pass filtered at 4 Hz. Principal component analysis (PCA) and hierarchical clustering analyses (HCA) were used to study EMG activity across the different movements and subjects.

RESULTS & DISCUSSION

We found a major difference in the number of principal components (PCs) required to explain 90% of the variance in the EMG data between the amputee (5 PCs) and able-bodied subjects (10- 11 PCs). In addition, HCA clustered the movement trials into four major subgroups consisting of wrist flexion/extension, wrist pronation/supination, wrist adduction/abduction, and all fingers based on all 10 subjects' EMG activity patterns.

The differences in the number of PCs between able-bodied subjects and the amputee could potentially be explained by the reduced muscle set in amputees, challenges related to muscle targeting, or more interestingly, changes in the ability to voluntarily make certain movements as a result of the chronic limb loss. The HCA results can be used to help visualize and understand the underlying patterns of EMG activity. The results of this study can be used to inform the design of bio-inspired controllers that generate prosthesis control signals from the biomechanical function of the muscles and the resulting movement dynamics.

THE UTILIZATION OF PATTERN RECOGNITION CONTROL FOR THE TRANSHUMERAL AMPUTEE WITHOUT TMR SURGERY: CLINICAL EXPERIENCES

Craig Jackman and Jason Macedonia

Hanger Clinic

ABSTRACT

Pattern recognition control has been commercially available since 2013 when COAPT released its Complete Control system. At the time of its launch, the clinical perception of the candidate selection for this control option was focused on those individuals who were proximal level amputees with TMR surgery. While it has been well documented that individuals with TMR surgery preferred pattern recognition control over direct control (1), the use of pattern recognition with non TMR proximal level amputees is yet to be definitively studied with commercially available prostheses. This presentation will share the author's experiences using COAPT's Complete Control system with two non TMR transhumeral amputees. Both of these individuals were previously fit with traditional direct control prostheses. In sharing these experiences, it will include the successes and challenges encountered in fitting these individuals with their prostheses as well as the perceptions of the users when comparing direct control with pattern recognition. The hope in sharing these cases is to inspire further investigation into utilizing pattern recognition for this population where an inclusion criteria can be established for pursing this technology.

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DESIGN OF A LOW-COST PROSTHETIC HAND FOR USE IN DEVELOPING COUNTRIES

Ashley Ballanger, Bsc; Ed Biden, D.Phil; Jon Sensinger, Ph.D.

Institute of Biomedical Engineering, University of New Brunswick

ABSTRACT

The World Health Organization (WHO) estimates that there are 25.5 million people with an amputation in developing countries who are living without any type of prosthesis [1]. Even with a lower incidence of upper limb loss than lower limb, there are likely several million people who could benefit from affordable, accessible upper limb prostheses.

When upper limb prostheses are available, users are typically provided a cosmetic hand or a body powered hook. Although a cosmetic hand provides the natural appearance that is often desired by users in less developed countries, it may not allow users to complete all activities of daily living (ADL). Conversely, a body powered hook is technically functional, but users are often uncomfortable with the appearance of the device. A third type of prosthesis, a body powered hand, is rarely used by people with upper limb deficiencies.

Body powered hands have the potential to provide a functional, aesthetically pleasing, and low-cost option to people in need of upper limb prostheses, but current designs are subjected to the highest rates of rejection of all terminal devices. Users have cited a variety of reasons for rejecting body powered hands [2]. At least two of these reasons, high activation force and low pinch force, can be attributed to mechanical inefficiencies in the device [3]. Existing body powered designs have been unable to decouple the actuation and the posture of the hand, leading to devices that actuate too many fingers and have poor efficiency, or actuate only one finger and have a poor selection of postures.

The authors have developed a body powered hand design which combines a single actuation point (the thumb) with the ability to independently pre-position the fingers and thumb. By only actuating the thumb, the device should require less energy to operate than currently available devices. The device is still capable of producing multiple hand postures, including tripod, lateral, and hook grasps, which should allow users to complete many ADLs. In this talk, the authors will present their design, along with data from mechanical and functional tests that compare performance of the prototype to currently available devices.

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MACHINE LEARNING TO IMPROVE PATTERN RECOGNITION CONTROL OF UPPER-LIMB MYOELECTRIC PROSTHESES

Frank Cummins[1], Aimee Feuser[1], Nathan Brantly[1], Levi Hargrove[1,2], Blair Lock[1]

[1]*Coapt, LLC,* [2]*Rehabilitation Institute of Chicago*

ABSTRACT

The clinical application of machine learning to prosthesis control is becoming better understood and more widely accepted. Commercially available pattern recognition systems employ machine learning algorithms to allow users to control their powered prostheses more intuitively, using their unique patterns of electromyography (EMG) signals. As users wear their devices more, the EMG signals they elicit for device control become more consistent [1]. There are factors, however, that can lead to changes in the characteristics of the EMG signals, which serve as inputs to the pattern recognition controller, such as electrode shift, muscle fatigue, et cetera. Currently, no commercially available pattern recognition system makes use of machine learning, supervised-adaptation algorithms to improve control via the utilization of historical EMG data collected during previous calibration routines. This paper introduces clinically relevant approaches and implementations of adaptive machine learning for control of prosthetic and orthotic devices.

INTRODUCTION

More than 11,000 amputations at the wrist disarticulation or higher-level occurred in the US between 2005 and 2013 [2] – with many more occurring worldwide. Pattern recognition control for myoelectric prostheses has benefitted many individuals with upper limb loss and limb difference since commercialization in late 2013 [3]. Described as “the single biggest breakthrough in [Pattern Recognition] in decades” [4], prosthesis-guided calibration played a significant role in the commercialization of pattern recognition control [5-8]. Researchers, clinicians, and users alike have expressed appreciation for the on-the-go recalibration feature, which puts the user in control of calibrating (i.e., updating) the pattern recognition system whenever and wherever desired – and without the need for a computer. However, despite the accessibility of the calibration scheme, it remains somewhat rigid: requiring the user to perform a sequence of all available prosthesis movements for each calibration.

All human-computer interfaces inherently involve two systems capable of adaptation: the human and the computer (i.e., the algorithm), with both systems affecting performance [1]. He et al. found a statistically significant trend of improved muscle contraction repeatability across days of use with a training paradigm that did not employ any external feedback. As the electrode locations and the electrode-skin impedance were controlled factors in the study, the increasing repeatability trend can be best explained by physiological adaptations of the subjects through learning to perform consistent muscle contractions. In addition and complementary to user adaptation, it is important that the pattern recognition control system (i.e., the algorithm) adapt to EMG signal non-stationarities, such as electrode location shift, muscle fatigue, and varying limb orientations during training. Furthermore, Vidovic et al. found that classification accuracy increased from 75% to above 92% when utilizing an adaptive calibration method as compared with a static training paradigm, a promising result for clinical implementation [9].

Some users choose to calibrate their prosthesis control multiple times each day, and calibration routine improvements are some of the most requested enhancements of the commercial pattern recognition control system. Currently available pattern recognition control systems discard collected EMG data by default when a new calibration is performed. The goals of the adaptive calibration approach are to improve prosthesis control (by utilizing a larger set of EMG data for training the pattern recognition control system) and to reduce the amount of time spent recalibrating the system. This can be accomplished by making use of the historical EMG data to improve the generalizability of the controller to EMG signal variability and adapting the controller with the most recently collected set of calibration data. In this contribution, a machine learning supervised adaptation calibration paradigm for improving prosthesis control and potentially reducing the need for recalibration is presented.

METHODS

Seven intact-limb subjects (four males and three females) and four subjects with transradial limb difference (three males and one female) completed the following IRB-approved experiment. An elastic cuff with eight, equidistantly-spaced, bipolar electrode pairs was donned on the upper forearm approximately two cm distal to the elbow with a ground electrode placed collinear with the olecranon. A software interface guided the collection of the following muscle contraction data: wrist supination and pronation, hand open, key grip, chuck grip, fine pinch grip, and point grip. These movements are commercially available and routinely used in powered prostheses (i.e., wrist rotators and multi-articulating hands). Subjects were verbally instructed to perform medium strength, constant-force muscle contractions but were provided with no biofeedback. Subjects completed seven data collection sessions each consisting of eight repetitions of all collected muscle contractions. This data collection procedure is common in the field of myoelectric pattern recognition control [1, 10-11].

Five paradigms for the training and testing of the pattern recognition classifier were examined: training on the first session and testing on each subsequent session ("Static", i.e., across-session testing with a static decoder), training on session N and testing on session N+1 ("Across", i.e., across-session testing), training with adaptation that remembers all data ("Pooled", i.e., training with data from all sessions prior to the testing session), training and testing with adaptation with a fixed memory ("Adapt.", i.e., across-session testing with adaptation), and training and testing with data from a single session ("Within", i.e., within-session testing). Classification error rates were used to assess offline classifier-training paradigm performance. A 2-way ANOVA with classification error rate as the response variable and training paradigm as a fixed factor was completed. Additionally, the impact of increasing the number of available functional hand grasps (i.e., one, two, three, or four hand grasp patterns) on classifier error rate was also examined. Finally, qualitative user feedback from beta-testing of the calibration scheme was analysed to assess clinical implementation.

RESULTS & DISCUSSION

Both subject groups (i.e., intact-limb subjects and subjects with transradial limb difference) show the same classifier-training paradigm performance trends. The results highlight that the adaptive calibration scheme resulted in classification error rates lower than the "Static" classifier-training paradigm, which performed the poorest of all five conditions (Figure 1). When compared with static decoding, the classification error rate of the supervised adaptation paradigm was significantly lower ($p < 0.05$). Presumably, the "Static" condition performed poorly because it could not adapt to EMG signals changes across the eight testing sessions, which mirrors how the existing commercial controller might behave if used on subsequent days without retraining. The "Within" condition, which mirrors how the existing commercial controller might behave if retrained before each period of use, performed best for both subject groups. The within-session testing condition was expected to have the highest classifier performance because the EMG forearm cuff was not doffed between the classifier training and testing, meaning the electrode locations and electrode-skin impedance values were effectively equivalent across the training and testing data sets. The results of the other conditions were not statistically different. Trends toward improvements in classification error rates were noted from the across-session condition to the across-session with pooled training data condition and further from "Pooled" to the across-session with adaptation condition. It is likely that more subjects are required to achieve appropriate statistical power to detect differences between the "Across", "Pooled", and "Adapt." conditions. The preliminary data support the hypothesis that the use of supervised adaptation would decrease the need for frequent recalibration of the pattern recognition control system. Further exploration to ensure clinical viability is necessary. It is expected that the performance of the adaptation classifier-training paradigm would approach and potentially surpass that of the within-session case with a greater number of data collection sessions added to the pattern recognition control system. Qualitatively, subjects found that prosthesis control improved and that controller recalibration was not needed as often with adaptive calibration.

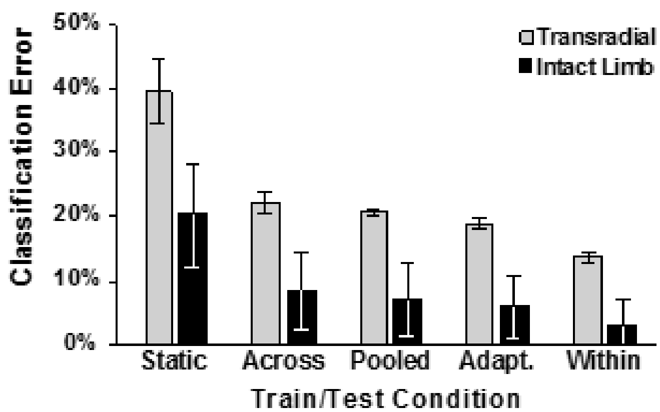

Figure 1: Results showing the classification error rates with standard error (error bars) for all five training paradigms arranged by subject group.

CONCLUSION

While many users express appreciation for the on-the-go recalibration (i.e., prosthesis-guided calibration) feature of commercial pattern recognition control, the rigidity of the calibration routine has at times proven to be burdensome. An adaptive approach may provide a means not only to reduce the frequency of recalibration, but also to improve functional prosthesis control. By adding new data to the classifier rather than completely clearing the classifier, we should develop a system that generalizes to more movements and use conditions. Further investigation into the robustness of adaptive calibration across many different muscle contraction patterns and clinical settings is being explored.

SIGNIFICANCE

Pattern recognition control of upper-limb prostheses is growing in clinical acceptance. The implementation of supervised controller adaptation into the commercial pattern recognition system is expected to improve real-time, home-use performance and decrease the need for recalibration, a development with far-reaching clinic impact. Improvements to calibration, especially those resulting in greater prosthesis control, improve the viability of pattern recognition control in comparison with conventional amplitude-based control approaches.

DISCLOSURE

Dr. Levi Hargrove has a financial interest in Coapt, LLC.

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ISENS – PROGRESS AND PROSPECT OF A FULLY IMPLANTED SYSTEM FOR SENSORIMOTOR INTEGRATION

Dustin Tyler

Case Western Reserve University

ABSTRACT

Since May 2012, peripheral nerve cuff electrodes have been providing direct sensory feedback to individuals with upper extremity limb loss and in lower extremity limb loss since May 2016. In April of 2016 and February 2017, we added new subjects with eight bi-polar myoelectric (EMG) channels for recording muscle activity for simultaneous, arbitrary multi-degree-of-freedom control in the upper extremity. There are a total of 48 nerve or EMG channels implanted on the peripheral nerves and in the muscles, but are routed to external stimulation and recording hardware via percutaneous leads. The sensory restoration system has provided stable feedback over multiple discrete points of the phantom hand since its implant; has eliminated phantom pain is subjects; can provide multiple qualities of sensation; and can provide the same capabilities of intensity discrimination as an intact hand. The implanted EMG electrodes have provided stable, isolated, high signal-to-noise recordings since implant and can provide naturalistic 3 degree-of-freedom control without retaining for over six months.

These results encouraged the development, starting in May 2015, of a clinically-viable, fully-implanted, wireless system to eliminate the percutaneous leads. The implanted somatosensory electrical neurostimulation and sensing (iSens) will drive 64 stimulation channels for restoration of sensory feedback and record from 32 channels, configured into 16 bipolar EMG recording electrodes, for device control. The system will connect to an external device via a high-reliability, low-power Bluetooth wireless link, The systems consists of a central battery and communications module (INC); four "smart leads" that are each connected to the INC via a four conductor lead and then connects to 32 stimulation or recording leads. The sensing smart lead can simultaneously record from 8 bipolar EMG channels with 10-bit resolution at 1000 Hz sampling rate. The stimulation smart lead can simultaneously stimulate twelve channels asynchronously at up to 100 Hz on each channel with patterned intensity stimulation paradigms. The final system will place 32 channels on the median nerve and 16 on each the radial and ulnar nerves. External transmission will be to a Bluetooth dongle connected to personal mobile device or lab laptop computer via USB that serves as the user interface and the main algorithm processor. The Full device engineering and component verification will be completed by end of 2017; full system verification, animal testing, and IDE submission by 8/18; and anticipated approval for clincial study initiation in early 2019.

MODULATION OF PHANTOM LIMB PAIN USING EPIDURAL STIMULATION OF THE CERVICAL DORSAL SPINAL CORD

Ameya Nanivadekar, Santosh Chandrasekaran, Ahmed Kashkoush, Eric Helm, Jennifer Collinger, Robert Gaunt and Lee Fisher

University of Pittsburgh

INTRODUCTION

Pain is a common comorbidity of conditions such as peripheral nerve injury, substance-induced neuropathy, and trauma. Nearly 1.5 billion people worldwide suffer from chronic pain with the estimated cost of health care nearly $275 billion. The mechanisms of neuropathic pain are poorly understood and its evaluation in humans is complex because most stimuli required to induce neuropathic pain produce irreversible damage. Recent evidence suggests that the incidence of chronic phantom limb pain can be regulated by delivering sensory feedback that is relevant to the amputated limb. This study aims to determine whether cervical spinal root stimulation to elicit sensations localized to the amputated arm can also result in concomitant changes in PLP

METHODS

All procedures were approved by the University of Pittsburgh Institutional Review Board and the US Army Human Research Protection Office. Two study participants were implanted with three 8 or 16 contact spinal cord stimulation leads (Boston Scientific) in the lateral epidural space of the cervical spinal cord. Stimulation electrode, amplitude, frequency and pulse width were varied across trials. The location, intensity and modality of the evoked percepts was recorded. The intensity of PLP was recorded on a visual analog scale (VAS) after every stimulation trial. Additionally, the McGill Pain Questionnaire (MPQ) was administered on a weekly basis, and again one month following explantation. The leads were explanted after 2-4 weeks.

RESULTS

A total of 1,493 trials evoked localized sensations, of which 580 PLP episodes were reported (38.9%) at a mean intensity of 2.5 ± 1.9 on the VAS. For the 115 electrodes that evoked a sensation, stimulation amplitude and pulse width were related to the intensity and incidence of PLP respectively. Furthermore, a clinically significant (>5 points) reduction in PLP was observed on the MPQ in subject 1 (9 points) and subject 2 (8 points) at 1-month follow-up. Additionally, the effect of stimulation electrode location on PLP modulation as well as the correlation between the modality of stimulation evoked non-PLP sensation and the incidence of PLP is being explored.

CONCLUSION

This study suggests that stimulation amplitude and pulse width may modulate the intensity and frequency of a PLP episode. We further observed time-dependent PLP modulation such that the immediate post-stimulation phase was associated with increased PLP that may be coupled to a long-term reduction in PLP.

THE SOFTHAND PRO-H: A PROSTHETIC PLATFORM FOR WORK-ORIENTED APPLICATIONS

Sasha B. Godfrey[1], Cristina Piazza[2], Manuel G. Catalano[1,2], Matteo Rossi[1,2], Matteo Bianchi[2], Giorgio Grioli[1,2], and Antonio Bicchi[1,2]

1. Advanced Robotics Department, Istituto Italiano di Tecnologia, Genoa, Italy. 2. Research Center "E. Piaggio", Department of Information Engineering, University of Pisa, Pisa, Italy

ABSTRACT

Body-powered prostheses are typically favored for heavy-duty use and employed in hostile environments, such as farm or factory work. With adequate training and practice, individuals with limb loss can become proficient in the usage of body-powered hooks (BPH) to accomplish a wide variety of tasks. Despite this versatility, there are drawbacks to this type of technology [1]. In this work, we present a novel prosthetic platform to address three of the most common issues in the use of BPH: 1. The need to frequently change terminal devices to task-specific solutions. 2. The loading of the shoulder due to the use of the figure-of-nine harness (the typical control system of body-powered prosthetic systems for users with unilateral limb loss). 3. The lack of functional, anthropomorphic solutions for users who prefer body-powered prosthetic solutions.

In contrast, myoelectric prostheses (MPs) are externally powered and controlled by muscle activity in the residual limb. Further, unlike the body-powered devices described above, they are typically anthropomorphic but more fragile, costly, and heavy. The most advanced versions offer multiple grasp postures, but are more difficult to control. Indeed, controlling even single degree of freedom (DOF) MPs can be challenging for some individuals, either because of their physiology or because of environmental factors [2].

The Pisa/IIT SoftHand [3] is a 19 DOF anthropomorphic robotic hand that combines intuitiveness, adaptivity and robustness. The mechanical design is based on studies on human kinematic synergies and leverages underactuation to simplify control and imparts adaptability to the grasp pattern. A previous prosthetic implementation of the Pisa/IIT SoftHand, the myoelectrically-controlled SoftHand Pro, has been developed and is being tested with individuals with limb loss [4]. This work presents the possibility of applying the SoftHand technology to tackle the issues identified above.

For this reason, we examined the feasibility of combining the benefits of both body-powered and myoelectric prostheses in a hybrid solution focusing on work-oriented applications. This solution uses a shoulder harness to control an externally-powered anthropomorphic prosthetic hand. We started by analyzing the placement of the main moveable components of the prosthesis (motor, battery pack and electronics) on the terminal device, socket, or user's body. Eight potential configurations were selected as feasible solutions, depending on situational requirements. This work presents this analysis as well as one of these solutions, which has been implemented as a functional prototype, the body-controlled, servo-assisted SoftHand Pro-H and featured in the Cybathlon 2016 Powered Arm Prosthesis Race.

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EVALUATION OF DAILY USE AND FUNCTION OF CONVENTIONAL BODY-POWERED PROSTHESES AND CUSTOM VO/VC TERMINAL DEVICE

Ashley Swartz[1], Kristi Turner[1], Arun Jayaraman[1,2], Jon Sensinger[3]

Shirley Ryan AbilityLab[1], Northwestern University[2], University of New Brunswick[3]

ABSTRACT

The majority of persons with upper limb amputation use body-powered prostheses due to simplicity, robustness, and low cost. However, little is known concerning the daily use, function, and average force exertion for these body-powered devices. In addition, body-powered prosthesis users must choose between a voluntary-opening (VO) and voluntary-closing (VC) device. It is yet to be determined whether a device capable of both VO and VC would provide added benefit and function.

The two main objectives for this study were to quantify the actuation frequency and force exertion for body-powered prosthesis users, and to investigate the impact of a novel VO/VC terminal device capable of being used in both VO and VC modes [1]. Four subjects with a trans-radial amputation were recruited and were fit with an instrumented harness. This harness contained load cell electronics in-line with the Bowden cable and measured the force exerted by the user to actuate the device, as well as the frequency with which the device was used on a daily basis. We also sent the subjects home with the novel VO/VC device, using the same mechanism to track the number of times they used the device as well as how often they switched the device between VO and VC modes. Following the home trial portion, all subjects performed outcome measures of Box and Blocks, Jebsen-Taylor Hand Function Test, Southampton Hand Assessment Procedure (SHAP), and Assessment of Capacity for Myoelectric Control (ACMC) with both their conventional and VO/VC devices, in randomized order. Subjects were also asked to complete a qualitative survey concerning their experiences with the VO/VC device.

All subjects chose to use both modes of the VO/VC device during home use and during the outcome measures. Two subjects performed better at both the Box and Blocks and Jebsen-Taylor using the VO/VC device over their home device. One subject performed better on the SHAP using the VO/VC over their home device. Although the VO/VC device used in this study was experimental all subjects chose to switch modes during both their outcomes and in daily use. This suggests that devices capable of switching modes are useful. The qualitative feedback questionnaires identified room for improvement in the mechanism which could lead to improved outcomes and performance. VO/VC devices in general appear to be useful in daily life and warrant further research attention.

ACKNOWLEDGEMENTS

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BIOMIMETIC MODEL-BASED HAND CONTROL: PROGRESS AND CHALLENGES FOR MYOELECTRIC PROSTHETICS

[1]Robert Gaunt, [2]Sergiy Yakovenko, [2]Valeriya Gritsenko, [1]Lee Fisher, [1]Misagh Mansouri Boroujeni, [1]Carl Beringer, [2]Anton Sobinov, [2]Matthew Boots, [1]Jennifer Collinger and [1]Michael Boninger

[1]*University of Pittsburgh*
[2]*West Virginia University*

ABSTRACT

Commercial upper-limb prostheses are functionally constrained by the lack of suitable myoelectric control signals. Although dexterous prosthetic hands are becoming more available on the market, controlling these devices often requires non-intuitive methods, such as inertial measurement units placed on the feet, or more commonly, selection of a grasp pattern followed by actuation. These control methods have a key factor in common; they require non-intuitive signals to replicate what able-bodied people achieve effortlessly. More advanced approaches, typically based on pattern recognition algorithms, are being developed to provide more intuitive control. However, as we move towards simultaneous control of the wrist and individual fingers, even these approaches face challenges. Here, we propose an alternative method that leverages anatomical and physiological knowledge of muscle function and hand biomechanics to create a biomimetic musculoskeletal control system. This approach assumes that multi-channel, fully implanted myoelectric recording systems, currently under final evaluation and testing, will be available in the near future. Here, we describe the general framework for this biomimetic controller and highlight some of the progress, as well as challenges, in developing such a system.

All procedures were approved by the University of Pittsburgh Institutional Review Board and the US Army Human Research Protection Office. Nine able-bodied subjects and one transradial amputee were enrolled in the study. Intramuscular EMG data were recorded from 16 fine-wire electrodes placed in the extrinsic hand and wrist muscles under ultrasound guidance. EMG and kinematic data were collected during structured hand and wrist movements.

At present, we have implemented a system that enables real-time control of simulated or physical prosthetic hands. We use Hill-type muscle models (29 muscles) and forward dynamic stimulations in MuJoCo to convert muscle activations, estimated from EMG signals, to muscle force, then joint torque, and ultimately movement. For the 18 mechanical degrees-of-freedom (DOF) and 29 muscles, this can be achieved in less than 1 ms. These simulations are generally stable, although noise in EMG signals and limited modelling of muscle activation-contraction dynamics, may limit current performance. Simultaneous control of 3-4 DOF is routinely achieve, although maintaining static postures remains a challenge. We are currently comparing simulated kinematic outputs to ground truth kinematics from able-bodied subjects to identify and evaluate the impact of changing model parameters. We believe that this overall approach will eventually enable restoration of dexterous prosthetic hand movements by encapsulating the normal musculoskeletal dynamics within the model while limiting reliance on training data sets.

THE YALE MULTIGRASP PROSTHETIC HAND

Michael Leddy and Aaron Dollar

Yale University

ABSTRACT

The last decade has seen significant advancements in upper limb prosthetics, specifically in the myoelectric control and powered prosthetic hand fields. Notwithstanding the improvements in functionality and control of myoelectric prosthetic hands, upper-limb amputees continue to prefer body-powered terminal devices. These body-powered systems have a purely mechanical cable driven actuation scheme that is nominally paired with simple single-grasp terminal devices.

The Yale Multigrasp Prosthetic Hand bridges the gap between body-powered and electric hands. The Yale Hand, a novel body-powered terminal device, is a low-cost anthropomorphic prosthetic hand that incorporates the advantages of multiple grasp types seen in many myoelectric hands. Our body-powered system provides the benefit of proprioceptive force feedback when grasping, requires purely mechanical control, and improves on overall system robustness with no required electrical components. The Yale Hand has three grasp types: power, precision, and lateral grasp that the user can select with a simple movement of the thumb. A single body-powered cable drives all three of the hand’s grasps and a modified whiffletree allows the force distribution for each finger to vary depending on the grasp used. The design of the asymmetric whiffletree allows for decoupling and passive compliance in the fingers during grasping. The fingers utilize a pin MCP joint and flexure PIP joint to provide out of plane compliance and an underactuated grasp response. The hand is anthropomorphic, sized to the specifications of a 50th percentile female hand, and features a 3d printed or carbon fiber/epoxy foam chassis. Our novel prosthetic hand preserves the durability, reduced cost and weight, and proprioceptive feedback of a body-powered split hook while encompassing the multi-grasp functionality and aesthetic appeal of more complex robotic hands.

The functionality of the Yale Multigrasp Prosthetic Hand was evaluated through benchtop testing and a twelve-subject able-body study. One unilateral trans-radial amputee and one bilateral trans-radial amputee performed evaluation studies to determine the level of dexterity achieved with the hand. Results show comparable performance to existing commercially available terminal devices on both the Box and Blocks and Southampton Hand Assessment Protocol for the able-bodied and amputee subjects.

DEVELOPMENT OF A SIMULATED SENSORY MOTOR PROSTHESIS: A DEVICE TO RESEARCH PROSTHETIC SENSORY FEEDBACK USING ABLE-BODIED INDIVIDUALS

Tarvo G. Kuus[1], Michael R. Dawson[2, 3], Katherine Schoepp[3] Jason P. Carey[1], Jacqueline S. Hebert[2,3]

[1] Department of Mechanical Engineering, University of Alberta, Edmonton, Alberta, Canada;
[2]Glenrose Rehabilitation Hospital, Alberta Health Services, Edmonton, Alberta, Canada;
[3]Department of Physical Medicine and Rehabilitation, University of Alberta, Edmonton, Alberta, Canada

ABSTRACT

Sensory feedback is a desirable feature for prostheses; however, research studies are limited in scope by the relatively small proportion of persons with upper-limb amputation. This impedes our ability to study the effect that various forms of sensory feedback have on device control and function. A Simulated Sensory Motor Prosthesis (SSMP) was developed to allow able-bodied users to perform functional tasks similar to a transradial amputee using a prosthesis, with the addition of somatotopically matched mechanotactile haptic sensory feedback. The intent is to assess the impact of relevant prosthetic sensory feedback on functional task performance.

This paper reports the design and development of the SSMP, which mimics the function of a prosthetic device, while also providing optional mechanotactile feedback. The device passed through many rapid iterations using 3D modelling and 3D printing, combined with traditional manufacturing techniques. The control of the device is similar to traditional transradial myoelectric prostheses, and required the development of training and testing protocols for new users. Data from twelve participants was collected and preliminary results are presented. A standard training protocol was successful at improving skill level to allow performance of 4 functional tasks. Participants gave higher ratings for confidence in grip security with the sensory feedback, compared to without. Two of the four tasks showed lower error rates using the sensory feedback. The SSMP provides flexibility to test and iterate different feedback modalities and control strategies as a first-pass with able-bodied participants. This offers the potential to save significant clinical and amputee participant time.

INTRODUCTION

Sensory feedback is listed as a desirable future feature for powered prosthetic devices [1]. Various approaches to providing sensory feedback in upper limb prosthetics are being pursued [2], including somatotopically matched feedback [3]. However, it can be challenging to recruit large enough populations of upper-limb amputee participants to reach statistical significance when evaluating sensory feedback strategies. As a result, many studies are case-specific and can be difficult to reproduce. Moreover, the method of introducing feedback is generally unique between studies, further decreasing reproducibility. Researchers have previously used simulated prostheses with able-bodied participants in the study of motor control to give greater statistical power by increasing participant numbers [4, 5]. Our goal was to develop a Simulated Sensory Motor Prosthesis (SSMP) for able-bodied participants, to study the potential impact of prostheses with sensory feedback. The SSMP was developed to specifically investigate the effects of mechanotactile haptic sensory feedback during functional tasks. Herein, “simulated” refers to the device simulating how a prosthetic device functions; “motor” refers to the control of the terminal prosthetic device; and, “sensory” refers to the mechanotactile haptic sensory feedback given to the user. This paper describes the SSMP device design and development, training protocol, and initial report on use during functional task testing.

DESIGN OVERVIEW

The SSMP was designed to simulate as closely as possible the function of a transradial prosthetic device for use with able-bodied participants. Because the SSMP is controlled through myoelectric signals in the forearm of the user, the weight of the device had to be minimized to not

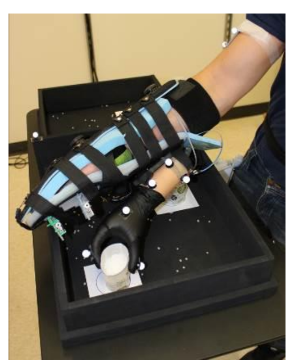

Figure 1: Side view of SSMP during use in cup transfer task

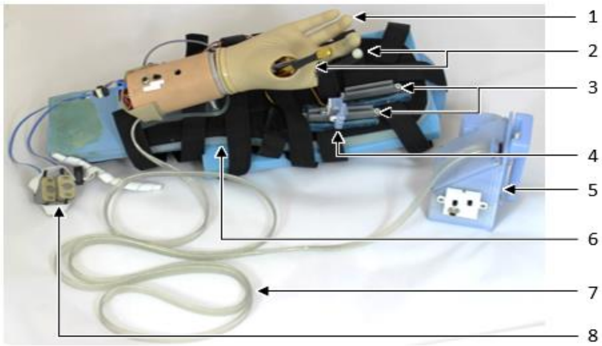

Figure 2: The SSMP with all required components, shown palmar side up, (1) sensorized Ottobock hand, (2) sensorized fingers with glove removed, (3) tracks for tactor adjustment, (4) tactor system, (5) electronics enclosure with belt clip, (6) hand brace, (7) cable connecting prosthetic hand to electronics enclosure, and (8) Ottobock electrodes.

fatigue the user's arm during use. The prosthetic prehensor was aligned under the palmar side of the user's hand with the fingers of the prehensor and the user as close as possible to each other. This was chosen to lower the moment arm from the elbow and still have intuitive and visible control (Figure 1). The forearm of the prehensor was also angled in the direction of the user's elbow to help the user associate with the device. The user's hand was restricted with the use of a rigid splint to ensure their hand did not move and would receive feedback at the same location throughout the testing protocol.

The device is controlled using myoelectric signals on the forearm of the user, with wrist flexion controlling prehensor hand close and wrist extension controlling hand open. Signals from two Ottobock electrodes (Electrode model: 13E200=60 Otto Bock Healthcare Products; Duderstadt, Germany) were fed directly to the Ottobock hand (MyoHand VariPlus Speed model: 8e38=9-R7 ¼ Otto Bock Healthcare Products; Duderstadt, Germany), where simple proportional dual site differential strategy was applied for control.

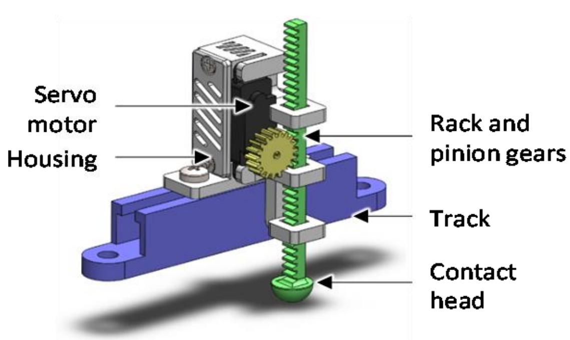

Figure 3: Tactor device within mounting track of the SSMP.

Most of the electronics and processors were placed within an electronics enclosure on the user's belt to reduce the weight acting on their arm. Electronic components were connected to each other via a durable flexible cable. The components of the SSMP are shown in Figure 2.

To measure the grasping forces of the prehensor, the fingers were fitted with strain gauge sensors (HDT Expeditionary Systems, Inc. a division of HDT Global; Solon, OH, USA,). Bending strain information from the sensors was converted to a force using the manufacturer's calibration. These forces were mapped to the tactors to provide matched feedback from the prosthetic fingers to the fingers of the able-bodied participant. The tactor system [3] (Figure 3) was driven using a HS-35HD Ultra Nano Servo (HITEC RCD; Poway, CA, USA); as forces are measured on the prosthetic hand, the servo motor rotates, causing the rack-and-pinion gear to push into contact with the user's fingertip. The device was untethered to improve ease of use during functional tasks, where settings could be manipulated wirelessly via Bluetooth.

MECHANICAL CONFIGURATION

To rapidly iterate SSMP designs, 3D modelling with SolidWorks 2016 and 3D printing were used. Specifically, the forearm and hand brace, prosthetic device attachment, electronics enclosure, tactor system, and strapping system were rapidly prototyped using PLA and flexible polyurethane filament. The final forearm and hand brace, as well as prosthetic device attachment were manufactured using polypropylene heated and shaped to the bottom of the brace and firm EVA foam for the hand brace. The rigid splint used a series of straps and Boa laces (Boa Technology, Inc.; Denver, CO, USA) which were attached to the brace to secure the user's hand and forearm.

A quick disconnect wrist allowed for the use of various prosthetic hands and provided adjustability in the rotation of the prehensor. We chose to provide mechanotactile haptic feedback to the pad of the fingertip on the user's thumb and index finger, matched to the instrumented thumb and index fingers on the sensorized prosthetic hand, to imitate ideal somatotopic feedback locations. To account for variance in user's hand sizes, slots were cut on the palmar side of the brace and a track for the tactor was mounted next to the slots to allow for alignment of the position of the tactors to the fingertip. Slot sizes and brace size were determined based on NASA anthropometric data [6] and accommodate users across the 5th to 95th percentile of the population. This meant the SSMP accommodated different sizes of forearms so that multiple users could be tested using the same device.

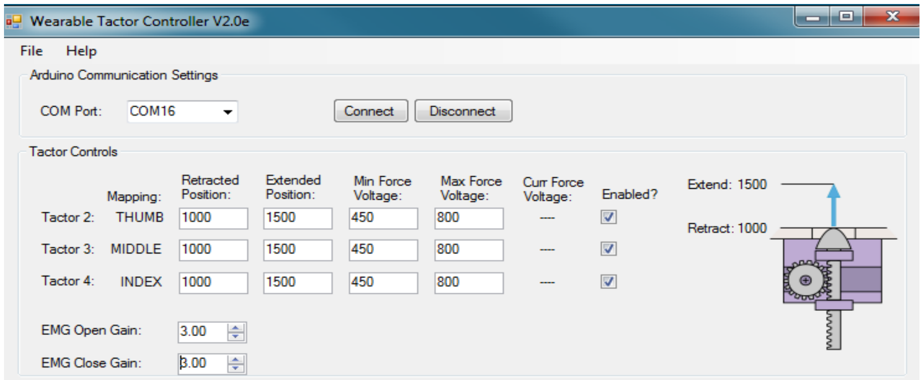

Figure 4: Wearable Tactor GUI

Straps were adjustable to account for different hand sizes and tactor positions. This was accomplished by lining the entire open area of the palmar side of the brace with hook and loop. The straps, once placed, could then be tightened using the Boa laces (by Boa Technology, Inc.; Denver, CO, USA). In previous iterations, discomfort was found to limit perception of sensory feedback. To improve comfort for the user, the inside of the brace was lined in PPT (patient protective technology) foam and polyurethane, foam (McMaster Carr; Aurora, OH, USA).

SOFTWARE DESIGN

A graphical user interface (GUI) was developed to change settings wirelessly and save the settings for each user (Figure 4). Each tactor can be enabled individually. Real-time force voltage is displayed in the GUI. The ranges of force voltage used for mapping can be adjusted, as well as the retracted and extended positions of the tactors.

The GUI does not need to be used for the system to operate, if adjustment of settings is not required. The overall time delay between force inputs to tactor output was quantified on a similar system to be less than approximately 150 ms. Users did not report a noticeable delay in the system.

The SSMP was required to measure sensory feedback even at low forces, i.e. when grabbing something very compliant, such as a wax cup. To enable the user to feel the sensory feedback with an object with little stiffness, a nonlinear mapping from measured force to delivered force was used (Figure 5). We used a series of linear maps with the greatest change (or sensitivity) in tactor movement at the lower end of the measured force, and less sensitivity at higher measured force.

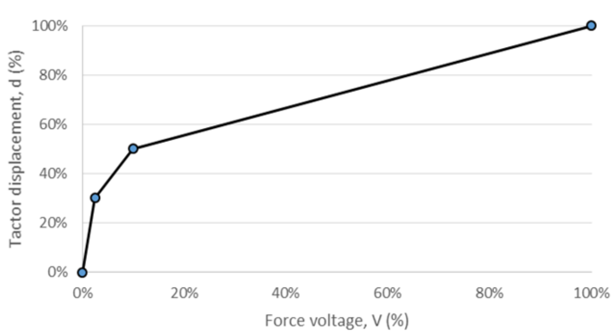

Figure 5: Tactor mapping trend line.

As both the index and middle finger of the prehensor measured force and the fingers moved in unison, it was reasonable to assume either or both fingers of the prehensor would contact an object during grasp. Therefore, the greater measured force from either the prehensor index or middle finger was mapped to the displacement of the tactor providing feedback to the user's index finger.

TRAINING AND TESTING

While most prosthesis users have substantial experience in operating their devices, this training is lacking in able-bodied participants. To ensure comparable results, it was desirable to train SSMP users to be able to reliably perform tasks and prevent a learning curve between trials from compromising the analysis. We developed a training protocol for the SSMP which included a myoelectric control strategy session, a general use of the SSMP guide, a

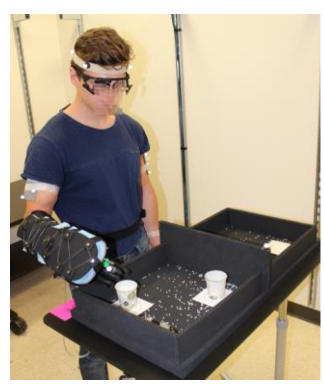

Figure 6: SSMP in use during cup transfer task, with the motion capture and eye tracking setup.

functional task training with and without sensory feedback, an object stiffness test, and an applied feedback location discrimination test. The total time for the training was approximately 1-2 hours.

Testing of the device included functional tasks evaluated using a 3D motion capture and eye tracking analyses, and a questionnaire. Four tasks were specifically designed to test the impact of sensory feedback. The tasks included were a pasta box transfer task, a cup transfer task (Figure 6), a cup pouring task and a shape sorting task. During training and testing short breaks were taken between trials and longer breaks were given if the user felt fatigued.

The total time for functional task testing was approximately 2-3 hours. This time included 3D motion capture and eye tracking measurement setup.

OUTCOMES TO DATE

The SSMP was used to test twelve able bodied subjects. The subjects underwent one setup and training session between 1.5 and 2 hours in duration, with a subsequent testing session using the protocol described above between 2 and 3 hours in duration.

At the end of each training session, participants were administered a questionnaire, and asked to rate their perception of using the SSMP with and without sensory feedback. In general, participants gave higher ratings for confidence in grip security with the sensory feedback, compared to without.

After the testing session, participants were asked to rank the difficulty and realism of the tasks performed. The tasks involving the wax cups (cup transfer and cup pouring) were rated as the hardest. This indicates that fine grip control is difficult, as excess grip would squeeze the cups and spill their contents, detrimental to task performance. For both the shape sorting and pasta box tasks, excess gripping force does not change task performance, so they were perceived as easier to perform. Furthermore, the wax cup tasks were also rated relatively high in terms of realism, meaning they represent tasks users would likely perform in daily life. The highest rated task for realism was the cup transfer task. Overall, there was no difference in average error between sensory feedback and none, but there were task specific differences. The pasta and cup transfer task both showed less errors with sensory feedback, but the easiest task (shape sorting) showed no difference and the most difficult task (cup pouring) had greater errors with sensory feedback.

CONCLUSIONS AND FUTURE WORK

An SSMP was designed and used successfully with twelve able-bodied participants to evaluate the impact of providing mechanotactile haptic sensory feedback on simulated prosthetic function. These trials have inspired several future design changes. The device should have more advanced data filters and an alternate method of mapping measured force to applied feedback, to reduce noise and relay feedback more effectively. There is potential to evaluate different modes of haptic sensory feedback using this device; for example, vibratory feedback as opposed to mechanotactile stimulus. Future revisions could provide access to more digits of the hand and possibly other areas of the arm, to evaluate alternate feedback locations. The device should also include grasping aperture measurement and data logging, with the potential to provide more effective feedback and control options.

Testing demonstrated that participants can successfully use the SSMP to perform functional tasks. Initial evaluation suggested that sensory feedback improved their confidence and performance for some tasks. Future work will involve a statistical comparison of the SSMP user data collected during the testing sessions compared to prosthetic users performing the same tests, to determine the similarities and differences in movement strategies and perception. Further comparisons of performance using the SSMP with and without sensory feedback will also be undertaken.

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MEASURE OF PAIN VARIABLES AND PRIMARY PROSTHESIS (BODY VS ELECTRIC)

[1]Tiffany Ryan, [1]John Miguelez and [2]Nathan Kerns

[1]Advanced Arm Dynamics
[2]University of North Texas

BACKGROUND

Numbness and phantom limb pain(PLP) have been found to negatively impact the upper limb(UL) amputees' functional ability and/or participation in activity. Clinicians must evaluate and monitor the impact of any association between prosthesis use and physical discomfort. Secondary to clinical experiences indicating a differences in physical discomfort severity between body powered(BP) and electric users, objective clinical survey results of UL amputees' report of residual limb numbness and pain, PLP and level of wear are presented.

METHODS

Consenting subjects from a convenience sample from patients presenting in seven out-patient UL prosthetic rehabilitation specialty centers completed the Comprehensive Arm Prosthesis and Rehabilitation Outcome Questionnaire-Revised©, (CAPROQ-R©) at various prosthetic fitting phases of care. These subjects objectively ranked the physical and functional factors influencing prosthetic performance. Categories reviewed for this study include: residual limb numbness and pain, phantom limb pain(PLP) and prosthesis wear time. The consented final sample size, after excluding non-responses, was one hundred eighteen. Results of subjects' survey most distant from the initial fitting were evaluated for this study.

A series of paired-samples t-tests were conducted to assess change in self-reported numbness, residual limb pain, and PLP when the participant was wearing their prosthesis versus not wearing their prosthesis. Analyses were conducted separately for participant that nominated a BP prosthesis as their "primary prosthesis" ($n = 27$; Mage = 46.61; 85.2% male) and those who nominated an electrically-powered prosthesis as their primary prosthesis ($n = 64$; Mage = 59.12; 70.3% male).

RESULTS

Results of the analyses of BP users found no significant change in numbness ($p = .858$), residual limb pain ($p = .340$), or PLP ($p = .826$). Similarly, results of the analyses of electrically-powered users found no significant change in numbness ($p = .780$) or residual limb pain ($p = .294$). However, there was a significant change in electrically powered users' PLP ($p < .001$); more specifically, these participants reported significantly less PLP when wearing their prosthesis ($M = 3.86$, $SD = 2.96$) than when they were not wearing their prosthesis ($M = 5.41$, $SD = 2.44$).

CONCLUSION

Objective outcomes describing the impact of prosthesis control on factors known to limit an UL amputee's activity engagement have the potential to positively influence clinical protocols and industry research and development. Continued research to further evaluate the impact of the variances in electric prosthesis types and materials and methods of prosthesis-human integration and the potential positive impact on medical outcomes.

A COMPARISON OF TRAINING APPROACHES FOR PATTERN RECOGNITION BASED MYOELECTRIC CONTROL

Kadie Wright, Erik Scheme and Kevin Englehart

Institute of Biomedical Engineering, University of New Brunswick

ABSTRACT

Decades of advancements in the development of myoelectric signal processing techniques have made prosthetic devices an effective means of functional replacement for major upper limb amputees. One of the control approaches that has been widely researched in this field is pattern recognition (PR) based control using electromyography (EMG) signals, which has only recently become commercially available. One challenge to its widespread clinical adoption to this point may be due to the need for training of the PR controller, which requires appropriate collection of example data. Although the inclusion of confounding factors (such as varying limb position) in the training data has been shown to significantly improve the performance of the pattern recognition approach, little work has focused on how to actually elicit the training contractions themselves.

This work examined two existing training techniques that are currently being used in the field (ramp contractions, and velocity guided training), and introduces two new alternative training methods; position guided training and a hybrid position and velocity approach. The comparison of approaches was motivated by a desire to incorporate more dynamic motion into the training process, which may better reflect the actual use case than existing methods and be more intuitive for users. It was hypothesized that more relevant training data would result in improvements in real-time performance and usability in a virtual target acquisition task.

Fourteen able bodied subjects (10 male and 4 female, mean age 24 +/- 2.2 years) completed a Fitts' Law based usability study using controllers trained with each of the training methods. For each method, EMG data representative of five different motions (hand open, hand close, wrist pronation, wrist supination, and no motion) were recorded and used to train the controller, before completing 24 repetitions of the target acquisition task.

Comparison of real-time performance metrics showed no significant difference between the ramp, position and hybrid approaches. Velocity guided training, however, as used in the previously reported prosthesis guided training, obtained significantly better movement efficiency ($p<0.05$). No significant differences were found in the Fitts' law summary metric throughput. These results suggest that, although other training approaches may offer more intuitive training prompts, the currently employed velocity guided training more effectively informs the training of pattern recognition based myoelectric control. Future work will include consideration of cognitive load and motivation on the part of the user, in order to help form a more complete picture of training and usability.

FACTORS INFLUENCING LONG TERM PROSTHESIS USE

Dan Conyers, CPO, FAAOP, John Miguelez, CP, FAAOP and Nathan Kearns, BS

Advanced Arm Dynamics, Inc.

BACKGROUND

Experienced Upper Limb(UL) prosthetists regularly engage patients with a successful long term prosthesis wear and use history. Clinical concentration to quantify the primary contributing factors to this outcome is paramount to consistent clinical practice. Patient population specific factors such as amputation level, patients' prosthesis expectations, experience and preferred type and demographic characteristics must be considered when making clinical decisions. While, there are many opinions as to what qualifies as prosthetic rehabilitation "success" (e.g., active grasp, user satisfaction). Clinical observation demonstrates a variance in prosthesis wear time among some patients and may serve as a common marker among UL patients. As such, the primary aim of the current study is to better understand which, if any, condition-related or demographic variables influence long term UL prosthetic rehabilitation "success."

METHODS

Representative case studies demonstrating the scope and variety of factors impacting long term prosthesis wear and use were completed. These representatives' results were compared to prosthetic rehabilitation patient survey results to identify corresponding factors influencing outcomes. One hundred and eighteen patients the Comprehensive Arm Prosthesis and Rehabilitation Outcome Questionnaire-Revised© (CAPOQ-R©) as a standard of clinical care. A series of t-tests, univariate analyses of variance, and simple linear regression analyses were conducted to assess associations between a) key demographic and condition-related variables and b) prosthetic wear time were completed.

RESULTS

Comparison of representative case studies and survey results confirmed prosthetist expectations of prosthesis wear history and daily wear time. Analyses of survey results found no significant difference in prosthetic wear time based on biological sex, education level, age, current job status, area of the country, trauma type, or primary prosthesis type. However, results showed a significant difference based on the amount of time patients had their prosthesis [$t(105) = 2.46$, $p=.016$]. More specifically, patients who had their prosthesis for more than five years ($M=9.59$, $SD=4.93$) wore their prosthesis for significantly more hours on a daily basis than participants who had their prosthesis for less than five years ($M=7.21$, $SD=4.15$).

CONCLUSION

Results of the clinical observation and patient responses suggest UL amputees with long term experience utilizing a prosthesis wear their prosthesis for longer periods of time each day. Results do not support any general or patient population specific demographics as having a significant impact on prosthesis wear time. Further research is warranted to evaluate the impact of other factors with the potential to influence long term wear and use of an UL prosthesis.

THE PSYONIC COMPLIANT, SENSORIZED PROSTHETIC HAND

Aadeel Akhtar, Kyung Yun Choi, Jesse Cornman, Wenjun Sun

PSYONIC, Inc.

ABSTRACT

We present a compliant, sensorized prosthetic hand that enables both motor control and sensory feedback for people with upper limb amputations. The hand has six powered degrees of freedom, corresponding to flexion/extension in all five fingers and thumb rotation. The dimensions of the hand are at 50th percentile female anthropometry.

The fingers of the prosthesis were designed to be compliant and can withstand sharp impact forces applied from anterior, posterior, and lateral directions. We achieve compliance in the distal and proximal interphalangeal joints through the use of a flexible bone inside of a silicone outer structure. Worm gears provide non-backdrivability to decrease power consumption when gripping objects with constant high torque. The worm gears and motors are protected from environmental shock since the compliant joints prevent damage to the gears.

Pressure sensing is achieved through a flexible printed circuit board that houses three pressure sensors and can wrap around the proximal interphalangeal joint of the finger. The finger is able to detect pressure through the use of three MPL3115A2 barometric pressure sensors (Freescale, Austin, TX) mounted on the flexible PCB. We cast the sensors in silicone (Dragon Skin 20, Smooth-On, Macungie, PA) to turn them into sensitive contact pressure sensors. The three sensors are placed over common areas of contact (fingertip, finger pad, and lateral finger) when making power and lateral grasps. The pressure readings can easily be mapped to sensations provided through vibrotactile, electrotactile, or other sensory feedback interfaces.

The entire hand can be built for less than $1000. This low cost makes research and development of sensorimotor prosthetic hands more accessible to researchers worldwide, while also being affordable for people with amputations in developing nations. Furthermore, the hand can be easily integrated into standard sockets, facilitating long-term use and testing of sensorimotor capabilities.

PROSTHESIS INCORPORATION: AN OUTCOME METRIC TO ASSESS TOOL INCORPORATION OF A PROSTHETIC LIMB

Adam Wilson, Satinder Gill, Daniel Blustein, and Jon Sensinger

Institute of Biomedical Engineering, University of New Brunswick

ABSTRACT

As new types of feedback systems are realized for prosthetic limbs, it is important to be able to assess the impact of those feedback systems on the amputee's experience. The Prosthesis Incorporation (PIC) outcome measure evaluates an amputee's level of tool incorporation of a prosthesis with feedback. The PIC scoring is performed using a modified crossmodal congruency test [1], while parameters such as training time, feedback agency (trust of feedback), spatial congruency (feedback distance from expected location), and physiological correspondence (naturalness of feedback) are controlled or measured.

The PIC outcome measure can be administered in approximately 20 minutes and immediately provides a quantitative measure of tool incorporation. The test uses feedback pairs to measure the user's reaction time while the pairs of feedback (feedback from the prosthesis feedback system plus visual feedback) are presented in complementary fashion (congruent) vs conflicting fashion (incongruent). The score is calculated as the mean reaction time difference between congruent and incongruent stimuli over four sets of 64 trials. A further assessment can be administered by delaying the user's feedback while measuring the delay in their actions [2]. This additional testing provides a measure of the user's feedback agency.

Our results, on 60 able-bodied subjects using a bypass prosthesis and 6 subjects with an upper limb amputation, show a proportional relationship between training time and PIC score, a proportional relationship between feedback agency and PIC score, a proportional relationship between physiological correspondence and PIC score, and an inversely proportional relationship between PIC score and spatial congruency. Based on this information and the results of the administered tests, engineers and clinicians can adjust and tune the amputee's feedback system to provide a stronger sense of incorporation of their prosthetic limb if necessary.

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GIVING THEM A HAND: WEARING A MYOELECTRIC ELBOW-WRIST-HAND ORTHOSIS REDUCES UPPER EXTREMITY IMPAIRMENT IN CHRONIC STROKE

Heather Peters, Stephen Page and Andrew Persch

The Ohio State University

OBJECTIVE

To determine the effect of a portable, myoelectric elbow-wrist-hand orthosis (MEWHO) on paretic upper extremity (UE) impairment in chronic, stable, moderately impaired stroke survivors.

DESIGN

Observational cohort study.

SETTING

Outpatient rehabilitation clinic.

PARTICIPANTS

Stroke survivors exhibiting chronic, moderate, UE hemiparesis (N=18).

INTERVENTIONS

Subjects were administered a battery of outcome measures testing UE impairment, functional performance and gross manual dexterity. They then donned a fabricated MEWHO and were again tested on the same battery of measures while wearing the device.

MAIN OUTCOME MEASURES

Outcome measures included the UE section of the Fugl-Meyer Impairment Scale (UEFM), a battery of functional tasks and the Box and Block (BB) test.

RESULTS

Subjects exhibited significantly reduced UE impairment while wearing the MEWHO (FM: $t=8.56$, $P<.0001$) and increased quality in performing all functional tasks while wearing the MEWHO, with 3 subtasks showing significant increases (feeding [grasp]: $z=2.251$, $P=.024$; feeding [elbow]: $z=2.966$, $P=.003$; drinking [grasp]: $z=3.187$, $P=.001$). Additionally, subjects showed significant decreases in time taken to grasp a cup ($z=1.286$, $P=.016$) and increased gross manual dexterity while wearing a MEWHO (BB test: $z=3.42$, $P<.001$).

CONCLUSIONS

Results suggest that UE impairment is significantly and immediately reduced when donning a MEWHO, and these changes exceeded the UEFM's clinically important difference threshold. Further, utilization of a MEWHO significantly increased gross manual dexterity and performance of certain functional tasks.

SURFACE MYOELECTRIC SIGNAL ADJUSTMENT FOR UPPER LIMB PROSTHESIS CONTROL APPLYING RT SYSTEM

[1]Kengo Ohnishi, [1]Kimihiro Hayashi and [2]Jumpei Oba

[1]Tokyo Denki University
[2]Kobe Gakuin University

ABSTRACT

Myoelectric controlled prosthetic hand has advantage of generating larger gripping force than the muscular force of the residual limb. This characteristic benefits small children to conduct tasks more easily. The hindrance of fitting myoelectric hand to small children is caused by the low reliability of myoelectric control at introduction due to low-reproducibility of non-adjusted myoelectirc sensor signal. The dialogical adjustment of the sensor applied to the schoolable child is not promising to younger children where basic assumption of repetitive concentrated muscle activations is questionable.

To overcome this problem, we propose on applying a realtime pattern recognition method, RT System, to myoelectirc sensor signal to skip the initial gain adjustment of the sensor amplifier. Recognition Taguchi (RT) system which is a modified strategy of Mahalanobis-Taguchi System is a statistical process that numerically scales the similarity of the sampled data cluster with the model data cluster by calculating two characteristic parameter, Signal-to-Noise ratio and sensitivity. The root-mean-square is computed from the two parameters generated from the groups of model data cluster and sampled data cluster. Then the root-mean-square distribution of the model data cluster is used to evaluate the difference of the sampled data.

As a pilot experiment, a conventional 3-pole dry electrode with analog filter and amplifier was used to sample raw myoelectric signal at 3kHz. A 60Hz-notch filter and 5-to-500Hz band-pass filter was applied digitally as pretreatment. Forearm extensor myoelectric signals of 2 male subjects, in their twenties, were recorded at the most suitable point and the most degraded point for 15s, respectively. For the RT system processed signal, series data of the first 0.33s window after discarding the initial 3s data of recording was set as the model data cluster in each collected sample. Ten seconds of the remaining collected data was processed as sample data. The RT system processed signal was compared to the conventional full-wave rectification, RMS-smoothing, envelope processed signal. While the conventional filtered data did not have clear contrast in degraded condition, RT system processed signal had muscle activation signals to be 200 times larger potentials compared to the signal at resting condition.

The RT system processed signal had superior amplification even for a subject with a thick subcutaneous fat that caused difficulty of operating an on-the-market myoelectric hand with conventional myoelectric sensor. However, the subject and experimental conditions were limited and further experiments are needed, especially with toddler, after safety concerns are cleared.

HIGH-DENSITY EMG FOR SIMULTANEOUS MULTIPLE MYOSITE VISUALIZATION AND IDENTIFICATION FOR MYOELECTRIC PROSTHESIS FITTING

Damini Agarwal[1], Sapna Kumar[1], Rahul Kaliki[2], Nitish Thakor[1, 3]

1. *Department of Biomedical Engineering, Johns Hopkins University*
2. *Infinite Biomedical Technologies LLC*
3. *Singapore Institute for Neurotechnology, National University of Singapore*

ABSTRACT

Myosite location is one of the most important steps in the process of myoelectric prosthesis fitting. This is because identification of the best locations for electrode placement governs the quality of EMG signals and the subsequent performance of control algorithms [1]. The process requires precision, even for two-site based direct control systems that use antagonistic muscle groups. The current industry standard is to manually palpate the residual limb while the patient performs a contraction to identify broad areas of muscle movement and then to use a differential electrode system for finer identification of myosites [2]. Shifting an electrode even by <1 cm over the muscle causes significant changes in sEMG amplitude subsequently affecting the quality of control [3]. The time required as well as the reliability of this process is solely dependent on the skillset of the prosthetist, thus making it a highly specialized procedure.

New control strategies such as pattern recognition [4-5] use up to eight EMG sites for signal acquisition. In the case of above elbow patients, all eight of these electrodes need to be placed in specific locations on the residual limb to maximize information content of each channel. With these emerging control strategies, the problem of myosite identification becomes increasingly difficult over traditional two-site direct control systems. Thus, there is a significant need to improve upon the traditional brute force method of myosite location.

We have developed a novel flexible High-Density EMG [6-8] array to "image" a patient's residual limb prior to socket fabrication. This system generates muscle activity maps from 128 channels of simultaneously recorded monopolar EMG signals. The muscle activity maps provide a visual means of identifying all potential myosite locations for a given contraction. Moreover, by analyzing different muscle activity maps for different hand motions and contractions, it is possible to determine the most unique combination of sites that provide differentiable patterns for control.

Use of this HD-EMG interface allowed for optimized identification of eight myosites for pattern recognition-based prosthesis fitting of a patient with a transhumeral amputation. Interestingly, in this case study, we identified unique EMG sites where there was little visually identifiable movement of the residual limb. Such sites would likely be missed by traditional myosite selection methods. Using the selected sites, the patient was subsequently successfully fit with a pattern recognition prosthesis. Thus, HD-EMG is a valuable myosite visualization and identification tool that augments the prosthetists' skillset in myosite location.

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PHAM: PROSHETIC HAND ASSESSMENT MEASURE

Chris Hunt[1], Rahul Yerrabelli[1], Caitlin Clancy[2], Luke Osborn[1], Rahul Kaliki[1,3], Nitish Thakor[1,4]

[1]*Department of Biomedical Engineering, Johns Hopkins University, Baltimore, MD, USA*
[2]*Department of Mechanical Engineering, Johns Hopkins University, Baltimore, MD, USA*
[3]*Infinite Biomedical Technologies, LLC, Baltimore, MD, USA*
[4]*Singapore Institute for Neurotechnology, National University of Singapore*

ABSTRACT

Current methods of assessing the functionality of prosthesis systems are often qualitative in nature. As such, there are issues with evaluation consistency and difficulty scaling these assessments across patient populations. Here, we describe an alternative outcome measure, the Prosthetic Hand Assessment Measure (PHAM) which quantifies the performance of manipulation tasks using body kinematics. Task performance scores are composites of individual deviation metrics and allow for standardized comparison across patient populations. It is our hope that the PHAM may aid both engineers in prosthetic systems development and clinicians in patient functionality assessment.

INTRODUCTION

Despite recent advances in upper limb prostheses over the last decade, device abandonment rates remain high at 35% with impaired functionality primarily driving rejection [1]. These statistics suggest a disconnect between the traditional outcome measures used to validate prosthetic system efficacy and the actual utility gained from these systems. Current outcome measures often evaluate performance in a small activity envelope with minimal degrees of freedom (DOF). For example, the popular Box and Block Test [2] evaluates the performance of a single DOF, open / close, throughout a 53.7 × 25.4 × 8.5 cm^3 area directly in front of the subject's chest. Furthermore, the manipulated objects, cubes measuring 15.625 cm^3, are not representative of common objects amputees are likely to interact with. These limitations are addressed in part by more advanced assessments, such as the Southampton Hand Assessment Procedure (SHAP) [3] and the Clothespin Relocation Test (CRT) [4]; however, their respective areas of evaluation still remain limited. Exploring larger activation spaces has proven increasingly important as emerging control paradigms, such as pattern recognition, are significantly less reliable when used in untrained positions [5]. In this respect, it is important to evaluate the utility of a prosthesis over a broad activity envelope in order to accurately report its efficacy.

Furthermore, the vast majority of validated outcome measures are subjective in nature, relying on clinician or user feedback to quantify quality. For example, the Activities Measure for Upper Limb Amputees (AM-ULA) [6] includes a measure of "movement awkwardness" as judged by a trained evaluator. Similarly, the modified Disabilities of the Arm, Shoulder and Hand survey (QuickDASH) [7] is completely dependent on the experience of the observing clinician and the personal opinions of the amputee. The aforementioned qualitative measures are useful; however, because they are qualitative, it is difficult to translate that usefulness across the wider patient population. There is a need for a more comprehensive, quantitative outcome measure through which assessment correlates to utility. This paper describes the Prosthetic Hand Assessment Measure (PHAM) as an alternative outcome measure to aid in the quantitative functional assessment of prosthetic systems.

KEY REQUIREMENTS

With the primary goal of accurately assessing prosthesis utility, the PHAM must:

- require the completion of tasks that approximate real-world use cases;
- explore typical DOFs over a significant 3D volume;
- be quantitative in nature;
- be scalable to future technological advances;
- be clinically deployable.

The PHAM approximates real-world use cases by requiring multi-DOF manipulation of common geometric primitives, combining features from both the SHAP and CRT. While each manipulation currently only requires open / close and wrist rotation, the PHAM is scalable in that each protocol can easily be modified to require further DOFs, such as wrist flexion / extension or ulnar / radial deviation, with no extra hardware. Additionally, the activity envelope of the PHAM is large enough to robustly investigate the effect position variance has on task completion rate and quality. Finally, the PHAM uses kinematic information to quantitatively assess quality of motion. While traditional motion capture schemes use optical or magnetic tracking, these systems are expensive in both cost and space; they are not clinically feasible [8]. Since the PHAM is to be clinically deployable, a flexible, multimodal motion capture system is implemented instead.

SYSTEM ARCHITECTURE

Hardware

The PHAM frame is a windowpane structure made from 1.5” diameter PVC pipes. The height and width of the windowpane are adjustable, ensuring that a user of any height is able to reach all segments of the windowpane with his or her prosthesis. The windowpane can be decomposed into twelve segments, six vertical and six horizontal, each containing LED strips and a PHAM testing environment (Fig. 1a). With each testing environment, the PHAM evaluates four different grips: power, tripod, pinch, and key. These grips are necessary to manipulate the four geometric primitives: cylinder, prism, block, and card. Each grip-object pairing is designed to be mutually exclusive, wherein each grip can only manipulate a single object and each object can only be manipulated with a single grip. These grip-object pairings are each targeted to simulate activities of daily living (ADL). These relationships are outlined in Table I.

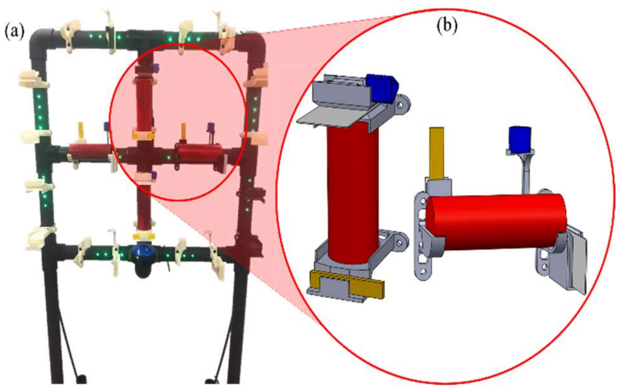


Figure 1: (a) An image of a complete PHAM windowpane. (b) Note that the horizontal and vertical testing environments are not the same. Slightly different designs are necessary to reduce the chance of collision while still guaranteeing each object can only be manipulated with one grip. The grip required to manipulate each primitive is represented by its color, the code of which can be found in Table I.

While similar to one another, the horizontal and vertical testing environments are slightly different in order to accommodate subtle, orientation-dependent requirements. While each testing environment houses a full set of geometric primitives, each object receptacle within the environment is not in the same relative position. This is to allow for proper hand clearance as well as reduce the chance of collision during task completion (Fig. 1b).

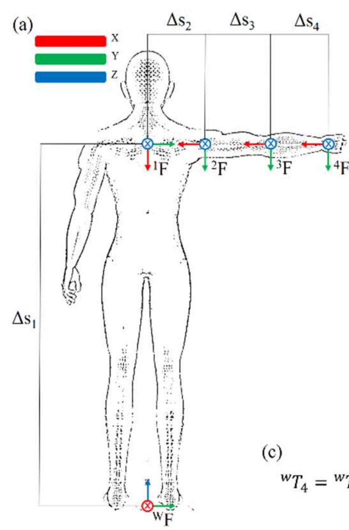
Table I: Geometric Primitives

Primitive	Grasp	Color	Activity of Daily Living
Cylinder	Power	Red	Pouring a Glass of Water
Prism	Tripod	Blue	Picking Up a Pencil
Block	Pinch	Yellow	Picking Up Coins
Card	Key	White	Grasping a Credit Card

Motion Capture

The PHAM system uses two sensing modalities in concert with one another in order to capture the dynamics of a subject during task completion: orientation tracking and force dispersion. Orientation tracking is accomplished through the use of five inertial measurement units (IMUs). Four IMUs are affixed to an anchor point on the subject’s body while the final IMU is located at the base of the PHAM unit, serving as a reference. With the method outlined in [9], the network of IMUs is able to return the orientation of each body segment of interest, including the subject’s trunk, upper arm, forearm, and hand. Using these orientations, normalized against the reference, it is possible to extract the corresponding joint angles, effectively reconstructing relative body position from frame to frame.

In order to maintain inter-subject consistency, ideal IMU anchor points were determined as such: the scapulae midpoint, the dorsal midpoint of the upper arm, the dorsal midpoint of the forearm, and the dorsal center of the hand. It is unlikely that each sensor will be placed in the same exact orientation from subject to subject and, so, a calibration routine is necessary to correct for sensor misalignment. For calibration purposes, ideal local reference frames are defined for each sensor and the transformation relating each adjacent frame can be found in Fig. 2. During calibration, the subject must attain the calibration position for one second, over which the average orientation is computed [10]. Once averaged, the apparent joint angles are calculated as the transformation relating each IMU’s orientation to that of its successor in the chain. The calibration angles are then simply the difference between the measured angles and the ideal transformations.



(b)

$$
{}^{w}T_1 = \begin{bmatrix} 0 & 0 & 1 & 0 \\ 0 & 1 & 0 & 0 \\ -1 & 0 & 0 & -\Delta s_1 \\ 0 & 0 & 0 & 1 \end{bmatrix}
\quad
{}^{1}T_2 = \begin{bmatrix} 0 & 1 & 0 & 0 \\ -1 & 0 & 0 & -\Delta s_2 \\ 0 & 0 & 1 & 0 \\ 0 & 0 & 0 & 1 \end{bmatrix}
$$

$$
{}^{2}T_3 = \begin{bmatrix} 1 & 0 & 0 & \Delta s_3 \\ 0 & 1 & 0 & 0 \\ 0 & 0 & 1 & 0 \\ 0 & 0 & 0 & 1 \end{bmatrix}
\quad
{}^{3}T_4 = \begin{bmatrix} 1 & 0 & 0 & \Delta s_4 \\ 0 & 1 & 0 & 0 \\ 0 & 0 & 1 & 0 \\ 0 & 0 & 0 & 1 \end{bmatrix}
$$

(c)

$$
{}^{w}T_4 = {}^{w}T_1\, {}^{w}T_1^{*}\, {}^{1}T_2\, {}^{1}T_2^{*}\, {}^{2}T_3\, {}^{2}T_3^{*}\, {}^{3}T_4\, {}^{3}T_4^{*}
$$

Figure 2: (a) The calibration pose with each IMU’s ideal reference frame defined. (b) The ideal rigid body transformation relating each adjacent reference frame, ${}^{B}T_{A}$. (c) Hand position, in the world reference frame, is given by chaining these transformations with the calibration transforms, ${}^{B}T^{*}_{A}$.

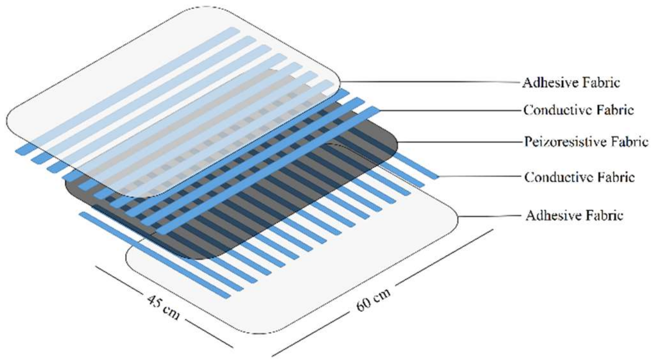


Figure 3: The mat is fabricated by adhering a peizoresistive fabric (black) between two orthogonal sets of conductive fabric strips (blue). When force is applied to a section of the grid, the resulting decrease in resistivity causes a change in voltage. The textile mat uses these voltage changes to track the 2D translational motion of the subject during assessment.

While the aforementioned modality is capable of capturing subject movement anchored at an immutable reference point, it is unable to capture the translational lower body movements often necessary to complete everyday tasks. As to not limit a subject's maneuverability during assessment, a simple translational motion capture subsystem is implemented using a force-sensitive mat. The mat is constructed using both conductive and peizoresistive fabric with the method illustrated in [11]. Using this design, the pressure matrix covers a 60 cm × 45 cm area with 150 discrete analog measuring units (Fig. 3). As the subject stands, the mat records the pressure distribution and locates the center of mass (CoM). All translational deviation in the XY-plane is computed as the movement of the CoM. Using this scheme with the IMU orientation tracking, it is possible to determine a subject's 3D kinematics during a task.

User Interface

The operation of the PHAM is controlled by a Python 3.5 GUI. After connecting all PHAM components, this user interface prompts the assessor to enter in experiment parameters, such as body segment lengths and task timeout. Before assessment, the user must choose a protocol, which determines the order and way in which the geometric primitives must be manipulated. A valid manipulation requires the movement of a single object from either a horizontal or vertical testing environment to a perpendicular testing environment. The PHAM supports three forms of protocols: preset, random, and custom. After the protocol is chosen and parameters are set, the PHAM LEDs flash orange to signify the start of a trial. This is an indication for the subject to press the large blue button to begin the first manipulation. After the initial button press, two PHAM segments light up, only one of which houses a geometric primitive. The subject will move the object from this lighted segment to the other lighted segment. The object to be moved is indicated by the color of the LEDs, outlined in Table I.

After this movement is completed, the subject must press the button again to indicate the end of that manipulation. In order to indicate a successful manipulation, all PHAM LEDs turn green. Immediately afterwards, two new PHAM segments light up, instructing the subject to complete the next manipulation. Should the subject take more than the allotted time to complete a manipulation, the PHAM lights will indicate failure by turning red and then turning off. Pressing the button will start the next manipulation. A total of four manipulations complete a trial for a preset or random protocol while this set size is arbitrarily large for a custom one. After a trial, the subject is allowed to rest while the assessor resets the positions of the PHAM objects in preparation for the subsequent trial. During all manipulations, regardless of completion status, the GUI records the kinematic data from the motion capture system.

PERFORMANCE EVALUATION

From the kinematics of a single manipulation task, several efficiency metrics are first computed. These metrics are defined as such: the 3D deviation of the subject's chest ($\overrightarrow{\delta_C}$), the 3D deviation of the subject's shoulder ($\overrightarrow{\delta_S}$), the 2D translational displacement ($\overrightarrow{\lambda}$), and the completion rate (η). The 3D deviation of any joint, $\overrightarrow{\delta_X}$, is defined as the summation of the absolute differences between 3D orientation in adjacent time intervals, seen in Equation (1).

$$\overrightarrow{\delta_X} = \begin{bmatrix} \delta_X^{\varphi} \\ \delta_X^{\theta} \\ \delta_X^{\psi} \end{bmatrix} = \sum_{n=1}^{N} \left| \begin{bmatrix} \varphi_X \\ \theta_X \\ \psi_X \end{bmatrix}_n - \begin{bmatrix} \varphi_X \\ \theta_X \\ \psi_X \end{bmatrix}_{n-1} \right| \tag{1}$$

2D translational displacement can be trivially calculated with the analogous method in two dimensions. Completion rate, η, is meant to most directly capture the utility of the prosthesis as the ratio successful tasks to attempted tasks. From these individual metrics, a succinct performance score, *P*, is computed as follows:

$$P = \frac{\left\| \overrightarrow{k^T \lambda} \right\|_1 + \left\| \overrightarrow{\delta_C} \right\|_1 + \left\| \overrightarrow{\delta_S} \right\|_1}{\eta} \tag{2}$$

Note that the L1 norm of $\overrightarrow{\lambda}$ is multiplied by the scaling vector *k,* proportional to the force mat dimensions. This is to balance the effect that units have on the final performance score. For assessment, a lower performance score equates to higher quality of motion and is more desirable.

RESULTS & DISCUSSION

To illustrate the quantitative analysis available through using the PHAM, a simple case study involving an able-bodied volunteer was conducted with their informed consent. The subject performed a custom PHAM protocol containing 16 object manipulations six times in total: three as a control and three again with their wrist braced. Performance for each manipulation is computed using the aforementioned metrics. The results can be seen in Table II.

Table II: Protocol Performance

Metrics		*Control*	*Braced*	*% Change*	*Units*
$\overrightarrow{\delta_c}$	φ	2.4849	2.3044	-7.26	rad
	θ	0.2418	0.3837	+58.68	rad
	ψ	0.9868	1.9303	+95.61	rad
$\overrightarrow{\delta_s}$	φ	1.3219	0.9122	-30.99	rad
	θ	0.2064	0.3091	+49.76	rad
	ψ	0.8840	0.9920	+12.22	rad
$\overrightarrow{k^T\lambda}$	X	1.6646	1.1469	-31.10	cm
	Y	2.1181	3.9590	+86.91	cm
η		1.000	0.875	-12.50	--
P		**9.9083**	**13.6429**	**+37.69**	**--**

The subject achieved a worse performance score with a wrist brace than they did on their control; however, the individual compensatory trends are not so black and white. While the reduced dexterity resulted in an overall increased L1 norm for every deviation metric, the composite deviations are not all positively trended. Unsurprisingly, the task completion rate, η, was significantly worse in the braced case than in the control. Interestingly, while the metrics displayed in Table II undoubtedly show increased compensation in the braced case, the 3D path travelled by the hand remains remarkably similar in both cases (Fig. 4). This suggests that while a reduction of DOFs results in an increase in compensatory movements, the redundancy of the human anatomy allows for near-optimal path tracking.

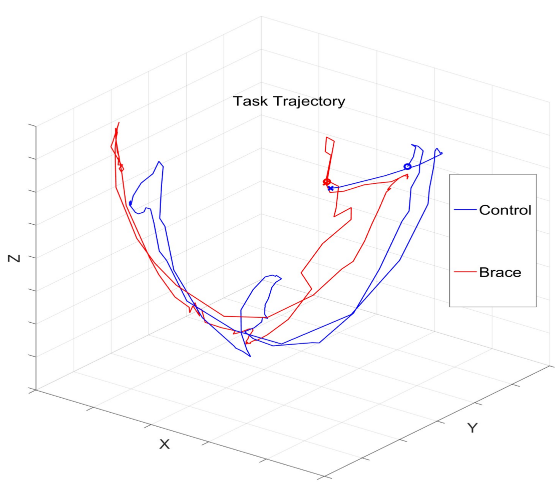


Figure 4: 3D hand path over a single manipulation in both the control (blue) and braced (red) condition for a single subject. Despite reduced DOFs in the braced case, both paths are quantitatively similar.

In this paper, we have introduced the PHAM as an alternative outcome measure used to quantitatively assess the efficacy of upper-limb prostheses. Furthermore, the system has shown that in-depth kinematic analysis is both possible and useful in a preliminary case study. Moving forward, we would like to compare the accuracy of the motion tracking subsystem to more traditional methods, such as optical or magnetic tracking. Additionally, we expect the kinematics of amputees to be significantly different even to braced, able-bodied subjects. As such, we need to validate the PHAM with amputee subjects in a large-scale study in the near future.

ACKNOWLEDGEMENTS

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CASE STUDY: BILATERAL ARM TRANSPLANT PATIENT AND USE OF PROSTHETIC DEVICES TO PROMOTE INDEPENDENCE AFTER TRANSPLANT

Joe Butkus and David Beachler

Walter Reed National Military Medical Center

ABSTRACT

The prevalence of arm transplantation, due to medical advances, has been increasing in recent years. Successful limb transplantation requires balancing of many issues, to include: extensive rehabilitation, medical management, financial support, availability of a caregiver and a tolerance for a decrease in functional abilities for some length of time. Patients agree to a period where they will have much less function than they were with prosthetic limbs that can last from 6 months to 18 months of limited function. After transplantation, it can be 6 months before a gentle functional pinch begins to emerge. This time commitment and extended period of decreased function complicates the patient's everyday life and the decision process for potential patients.

This presentation will examine the adaptations post transplantation and prosthetic options trialed to assist in activities of daily living for one transplant patient. Treatment course and collaboration between prosthetists and occupational therapists will be reviewed as the function of the hands changed as well as development of a prosthetic limb evolved. Other adaptations to the environment will be reviewed to educate participants in other ways to achieve success despite having hand or prosthetic control issues. Prosthetic options offer much faster path to a functional grasp and more intimate interaction with their environment.

The case review will serve to educate and illustrate issues and that arose during the first year of treatment. This review will help medical practitioners understand these options and find ways to promote greater independence at an earlier point in care. Patients can benefit from the return to some assistance from a prosthetic if they are willing to tolerate the time and effort involved. Achieving independence with all required daily tasks is the overarching goal and working to combine technologies can assist in that effort. These adaptations and combination of technologies will serve to improve the participants patient problem solving across a variety of other injuries.

QUANTIFYING MUSCLE CONTROL IN MYOELECTRIC TRAINING GAMES

Aaron Tabor[1,2], Wendy Hill[3], Scott Bateman[2], Erik Scheme[1]

Institute of Biomedical Engineering[1], University of New Brunswick
Human-Computer Interaction Lab[2], University of New Brunswick
Atlantic Clinic for Upper Limb Prosthetics[3]

ABSTRACT

Myoelectric training games have recently gained interest for increasing motivation and engagement when learning prosthetic control. However, game-based training has not yet been shown to result in improved performance of functional tasks, which has led to a push for "task-similar" training exercises and a questioning of the merit of training games altogether. This apparent lack of observable skill transfer remains counterintuitive, because games can encourage movements similar to those required for prosthesis control. To better understand the effects of game-based training, we identify a set of 'muscle-control metrics' to quantify characteristics of EMG control input that are considered important for 2-site proportional control. In this paper, we introduce these muscle-control metrics and describe a myoelectric training game developed in collaboration with patients and clinicians that is able to capture metrics during gameplay. We also outline an on-going data collection study, which will allow us to identify which aspects of a myoelectric training game lead to improvements in input signals.

INTRODUCTION

To take advantage of muscle plasticity and maximize potential for success, it is desirable for new myoelectric prosthesis users to begin training as early as possible following amputation (early childhood in the case of congenital limb-difference) [1]. However, a delay is often incurred before patients receive their prosthesis due to factors including recovery time, insurance processing, and fabrication and fitting of the prosthesis, and muscle training tools are used to keep patients active while waiting. However, these activities are often monotonous or lack sufficient feedback, making it difficult for patients to stay motivated [8].

Game-based training tools have been proposed to address the loss of motivation that patients often experience [3,4,5,7]. Muscle-controlled games can provide an engaging experience, making the otherwise monotonous training exercises more enjoyable. Therapists and prosthetists consider training games – and the muscle improvements they create – to be a valuable part of the training process. In practice, even simple games are used early in training to improve understanding, strength, and endurance, without the expectation of achieving functional skill transfer [7].

Despite their widespread use in practice, skills acquired in games have not been shown to transfer to functional prosthetic control, leading some researchers to question the benefit of training games altogether [3,4]. These studies, however, have only looked at coarse game performance (e.g., levels and score) as an indicator of learning, so the reasons *why* success in muscle-controlled training games does not predict improved functional control is still unclear.

To address the lack of information about the nature of improvement in training games and in myoelectric control, we propose a set of *muscle-control metrics* for assessing skill during the pre-prosthetic phase of myoelectric training. These metrics, inspired through conversations with clinicians, more accurately assess performance by quantifying aspects of muscle control that are important for success with a prosthesis. We built the ability to quantify and track these metrics into a training game we previously created through a user-centered design process with patients and clinicians [7].

Our work makes two main contributions: 1) we provide a set of objective and measurable metrics for tracking and assessing changes in muscle control; and, 2) we provide a freely available training game that enables the collection of these new muscle-control metrics. In the remainder of this paper, we describe work on game-based training for myoelectric control, introduce our muscle-control metrics, present our training game, and finally, outline our ongoing data collection that will allow us to better understand the nature of skills acquired through training games and how they might transfer to functional control.

BACKGROUND

Myoelectric Training Games

Recent research on training games has shown conflicting results. One study, employing the PAULA

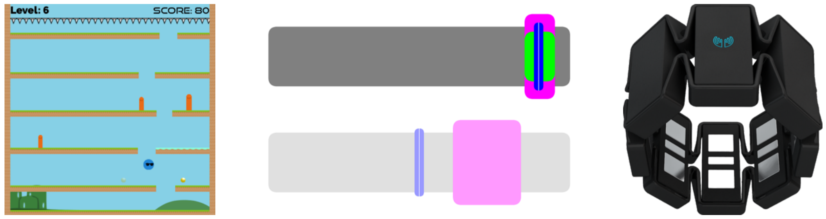

Figure 1: Research Tools. a) The Falling of Momo: Myoelectric Muscle-Training Game (github.com/hcilab/Momo.git), b) MyoFitts: Myoelectric 2-DOF Fitts Test (github.com/hcilab/MyoFitts.git), c) Myo Armband: Myoelectric Device (myo.com)

software suite, found that game-based tools are just as effective as traditional training approaches and that clinicians and patients can use games and other training activities interchangeably [2]. Conversely, follow-up studies by the same authors have suggested that some training games may be no more effective than a total lack of training. This recent work has suggested that, while patients might acquire skills in the game, they may not translate to improvements in prosthesis control [3,4]. These conflicting results suggest that the specific challenges associated with designing training games for amputees are not fully understood.

Outcome Measures

Assessing a patient's myoelectric ability is crucial for tracking progress over time. Existing measures are divided into two categories: 1) observational, and 2) self-reported [9]. *Observational measures* focus on quantitative, functional metrics such as task completion time, while *self-reported data* tend to focus on subjective elements such as perceived usefulness and embodiment of a prosthesis. Although an important focus during clinical training, muscle signal quality - an underlying prerequisite for robust myoelectric control – is often overlooked in existing observational metrics.

GAMES AND TRAINING ACTIVITIES

To support our research in training games, we have developed a game, called the *The Falling of Momo* (Figure 1a), and a standalone training activity based on acquiring targets, called *MyoFitts* (Figure 1b).

Momo is a muscle training game in which the player (blue) navigates their descent through a series of continually rising platforms [7]. Momo's design was aimed at creating a fun and engaging game that requires muscle movements that align with those required in prosthetic control. Flexion and extension contractions move Momo left and right, while a mode-switch co-contraction causes him to jump. Momo's features and design were informed and iteratively developed throughout a user-centered design process. Our process engaged amputee patients and clinicians through play testing and interviews, which we iteratively used to refine our game. Our work (described in [7]) proposes a set of design requirements for building training games that best meet the needs of patients and clinicians.

MyoFitts is a myoelectric targeting test with similar controls. Flexion and extension contractions moves the cursor (blue) into the targets (pink: un-acquired, green: acquired), with mode-switching being used to change focus between bars. Unlike Momo, MyoFitts is not a game, but instead is a training activity used in myoelectric control research (e.g., [6]).

Both Momo and MyoFitts are controlled with a 2-site proportional control strategy using the Thalmic Labs Myo Armband (Figure 1c), a commercially available myoelectric input device, which we have previously assessed as viable for use in training [7]. Both Momo and MyoFitts are freely available tools (see links in Figure 1) that incorporate data collection based on the muscle-control metrics outlined below.

MUSCLE-CONTROL METRICS

The following metrics are proposed for quantifying muscle control ability and are derived from logs of the EMG data captured during game play.

Muscle-Control Metrics

Several metrics are proposed to quantify common characteristics of muscle signals that are beneficial for all aspects of 2-site proportional control.

Isolation: When using a prosthesis with difference-based proportional control, greater muscle isolation enables a wider range of proportional speeds.

Isolation is computed as the ratio of intentional to unintentional muscle activity. It is calculated by inferring the intended direction of motion (i.e., stronger of the 2 EMG readings), and dividing by the level of unintentional co-contraction (i.e., the weaker of the 2

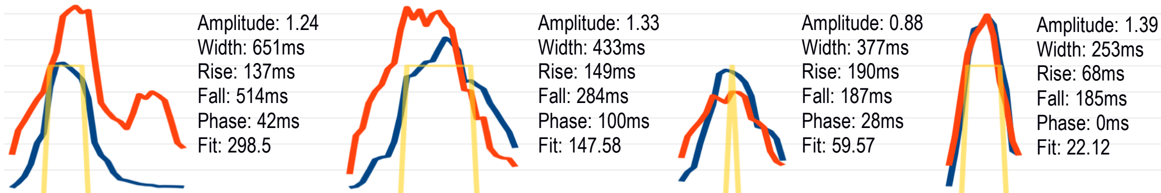

Figure 2: Co-contraction Improvements. Each co-contraction depicted was generated by a single participant during a series of 4 pilot training sessions (leftmost: session 1, rightmost: session 4). Red and blue lines represent flexion and extension signals, respectively, while yellow spikes indicate periods when the training system detected a mode-switch. Metric scores are shown to the upper-right.

EMG readings). Periods of impulse (described below) and rest (i.e., both readings below a threshold) are excluded. *Higher is better.*

Over-Exertion: Learning to create muscle contractions of an appropriate strength helps patients increase endurance while using a prosthesis.

Over-exertion is the weighted-tally of all EMG readings above the calibrated maximum voluntary contraction level, averaged by the number of samples. Higher values reflect unnecessarily strong contractions. Periods of rest are excluded. *Lower is better.*

Mode-Switch Metrics

Accurate co-contractions allow a prosthesis user to mode-switch reliably, limiting unintentional device movement. Mode-switching, however, is often a source of frustration, even for experienced prosthesis users [8]. Therefore, many of our metrics focus specifically on mode-switching and are calculated over brief periods of co-contraction. Co-contractions are detected by first identifying the registration time of each mode-switch, then searching forwards and backwards through the log for onset and conclusion times of the co-contraction, respectively (Figure 3, right). Figure 3 introduces terminology and demonstrates each of the metrics described below.

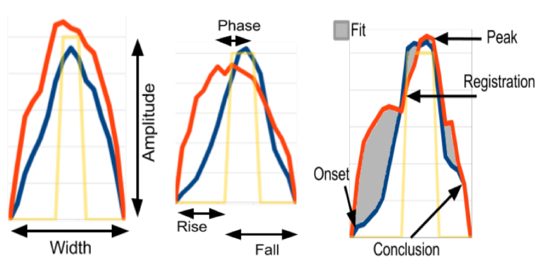

Figure 3: Terminology and Examples. Red and blue lines show the levels of flexion and extension muscle contraction, respectively. Yellow spikes indicate when the training system detected a mode-switch.

Amplitude: Mean height of flexion and extension signal peaks achieved during co-contraction. *Higher is better.*

Width: Duration of time between onset and conclusion of the co-contraction. *Shorter is better.*

Rise: Duration of time between onset and registration of the co-contraction. *Shorter is better.*

Fall: Duration of time between registration and conclusion of the co-contraction. *Shorter is better.*

Phase: Duration of time between the peaks of flexion and extension muscle signals during the co-contraction. *Shorter is better.*

Fit: Absolute difference in area between the flexion and extension signal curves during the co-contraction. *Smaller is better.*

In agreement with previous results [1,6], our pilot data suggests that game-based training can lead to improved muscle control and demonstrates how our metrics can be used to quantify and track these improvements (Figure 2).

DISCUSSION

Transfer to Improved Functional Performance

While our metrics are not yet clinically validated, they are based on principles of 2-site proportional control. Because of this, it is likely that they effectively characterize aspects of control not currently captured by coarser measures, such as scores in training games or completion times in functional assessment tasks.

We are currently running an experiment where new myoelectric users train using our game over a period of ten or more play sessions. In this experiment, we are tracking progress both in game play and through our muscle metrics, enabling a direct comparison between in-game learning and improvements in muscle control. Before participating in the study, all participants are briefed on study details and provide informed consent in agreement with the University of New Brunswick Research Ethics Board (REB 2017-047).

We believe that the metrics may also have further applications. Analyzing muscle-control improvement in detail allows clinicians to present patients with much more specific, targeted, and quantifiable training goals. Additionally, our metrics can be incorporated into games designed for at-home training and provide targeted feedback and increased awareness of progress made between clinical visits. Finally, our metrics are not specific to games; if EMG logs were collected from other activities (e.g., functional prosthetic tasks), the metrics obtained could be compared to those obtained during training games for discrepancies.

Limitations of the Current Muscle-Control Metrics

Some characteristics typically associated with strong myoelectric muscle control are not captured in our current metrics. Neither *consistency*, the ability to create and sustain a desired level of contraction strength, nor *endurance*, the ability to perform for prolonged periods of time without experiencing fatigue or performance degradation, are reflected in our metrics. Previous studies have assessed improvements in endurance by having participants perform a myoelectric tracking task both before and after training, recording the time at which participants became fatigued [6]. To our knowledge, improvements in consistency has not been assessed in the past.

When first learning myoelectric control, it is common for a patient to attempt a mode-switch several times before succeeding. Unsuccessful co-contractions (i.e., when the patient's attempt was not registered by the training system) hold valuable training information, but are not currently accounted for in our metrics. Incorporating these occurrences could help clinicians provide patients with even more constructive advice.

We observed in our pilots that, as participants got familiar with our training game, they began frequently "sliding into" and "sliding out of" mode-switches (i.e., quickly transitioning between left/right movement and jumps without relaxing muscles between the two phases). It is difficult to algorithmically distinguish between this "sliding" behavior and genuinely problematic impulses, so to avoid artificially inflating mode-switch metric scores we foresee the need to perform a more intelligent identification of mode-switches within EMG logs. It is interesting to question whether this "sliding" behavior is evidence of the development of a bad habit, or of a sense of proficiency and comfort with the control scheme.

CONCLUSION

In this work, we introduce a new set of metrics that are well suited to quantify improvement that occurs during myoelectric training games. We suggest that the information provided through these metrics is key to understanding skill transfer between training activities and functional control. Further, we believe our metrics would be beneficial to clinicians as they guide new patients through the training process, allowing them to identify specific deficits in control with greater precision. We have also demonstrated how the metrics can be employed and interpreted through their incorporation into a carefully designed training game, and a commonly used training activity. Our tools are freely available and may help provide critical new findings regarding skill transfer between game-based training activities and real-world prosthesis control.

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DERIVING PROPORTIONAL CONTROL FOR PATTERN RECOGNITION-BASED FORCE MYOGRAPHY

Alex Belyea, Kevin Englehart and Erik Scheme

Institute of Biomedical Engineering, University of New Brunswick

ABSTRACT

Force myography (FMG) has been proposed as an alternative to electromyography (EMG) for controlling a powered prosthesis. Previous research has varied in sensor type and configuration, and in control signal processing approaches. Some groups have used low numbers of sensors, while others have included high density grids (HD-FMG). These HD-FMG systems have been shown to reach offline classification accuracies as high as 99.7% for up to 8 classes of motion. As has been shown, however, high offline classification accuracy does not necessarily ensure a high level of prosthetic device usability. One large factor that contributes to the usability of a device, beyond its classification accuracy, is the use of proportional control.

As a precursor to an ongoing real time usability study, this work focussed on developing a proportional control scheme for use with pattern recognition based HD-FMG. Initial pilot work employed the mean of all channel outputs, as has long been used for proportional control in pattern recognition based EMG systems. It was found, however, that the HD-FMG signal did not monotonically increase with increasing effort. This is understandable given the subtleties of muscle synergies and their resulting patterns of mechanical deformation during graded contraction. Here, a class-specific proportional control signal was computed using a regression model trained to map FMG levels to the position of a prompted target.

In order to evaluate the performance of these approaches, HD-FMG and load cell data were collected from 14 participants as they matched their effort to a visual target prompt. Subjects were prompted by a visual position prompt to elicit 4 different active classes of wrist motion with their wrist constrained in a load cell device. From these data, the two proportional control were computed, and compared to both the load-cell value and the position of the visual prompt.

Overall, the average classification accuracy between the 5 motions was found to be 99.9%. The mean value proportional control approach yielded an R2 coefficient of determination of 0.453 with the visual prompt target. The regression based mapping approach resulted in an R2 coefficient of determination of 0.875.

This work represents a step towards real-time usability assessment of a HD-FMG based control scheme. These results suggest that a class-specific regression-based proportional control scheme may be effective for use as part of a pattern recognition based system.

FORCE SENSING PROSTHETIC FINGER TIP USING ELASTOMER-EMBEDDED COMMODITY INFRARED PROXIMITY SENSOR

[1]Jacob Segil, [1]Radhen Patel, [1]Yanyu Xiong, [1]Marie Schmitt, [2]Richard Weir and [1]Nikolaus Correll

[1]University of Colorado at Boulder

[2]Biomechatronics Development Lab. University of Colorado Denver Anschutz Medical Campus

ABSTRACT

The field of upper limb prosthetic design seeks to recreate what was lost after amputation including the loss of sensation. In order to accomplish this feat, prosthetic devices require compact, stable, and clinically robust sensors in order to measure interaction with the external environment. These types of sensors are more important now since dramatic progress was made to provide stable natural touch perception using implanted neural interfaces. We developed a force sensing prosthetic fingertip using elastomer-embedded commodity infrared proximity sensor to enable sensory restoration after upper limb amputation,.

The fingertip sensor integrates a commodity infrared proximity sensor (VCNL 4010, Vishay Semiconductors) which is embedded in a soft polymer, polydimethylsiloxane or PDMS (Dow Corning Sylgard 184). The infrared sensor was chosen due to its small form factor (3.95 x 3.95 x 0.75 mm3) which includes all digital electronics to produce a I2C communication output signal. The polymer was chosen due to its ease of manufacturing/molding and resistance to chemical and mechanical abrasion. The sensor operates by emitting infrared light (890nm) through the PDMS layer and measuring the net reflected intensity of the reflected signal. A layer of copper was deposited onto the PDMS material in order to reflect IR light and create a measurement which is independent of the reflectivity of the object in contact with the sensor. The intensity is approximately inverse of the square of the distance traveled and therefore can be used to determine displacement of the PDMS layer. Then, Hooke's law indicates the force and thereby creates a compact, stable, and clinically-robust force sensor for prosthetic fingers.

The mechanical design of the fingertip involved reverse engineering a commercially available prosthetic finger (Bebionic 2, RSL Steeper) into a computer-aided design (CAD) file. The CAD file was then modified in order to embed the printed circuit board of the sensor as well as pathway for the four-line ribbon cable. The fingertip was then manufactured using a plastic rapid prototyping printer (Objet Connex 350). Afterwards, a molding process created the PDMS layer which ensures that any contact force (i.e. - oblique or non-normal forces) will be detected by the sensor.

A force sensing prosthetic finger was developed in order to create a compact and stable measuring method to promote the restoration of tactile sensation for people with upper limb amputation.

MYOELECTRIC ELBOW-WRIST-HAND ORTHOSES (MEWHO) USED TO RESTORE FUNCTION IN A WEAK UPPER EXTREMITY RESULTING FROM CHRONIC STROKE - A CASE STUDY REPORT

Jonathan Naft, CPO, BSEE

Stefanie Dunaway, MS OTR/L

Myomo, Inc., Cambridge

ABSTRACT

Every year in the U.S. approximately 795,000 people, (civilians, active military and veterans) experience a stroke. Approximately 40% of that group exhibit chronic disability including upper extremity (UE) impairments such as paresis and spasticity. Additionally, within the military, 33,149 U.S. personnel were diagnosed with a TBI in 2011 alone. For those survivors who undergo traditional rehabilitative therapies, they may frequently be left with chronic upper extremity impairments and an associated loss of function, dependence on caregivers and decreased quality of life. Custom fabricated myoelectric orthoses may provide an alternative solution to improve function and an adjunct to traditional therapies for those with hemiparesis and loss of UE function. It is the aim of this case report to describe the experience and assistive and rehabilitative outcomes of a veteran with one such custom myoelectric orthosis.

The myoelectric brace used for this case study is called the MyoPro. The Myopro® (Myomo Inc. Cambridge MA) is a custom fabricated myoelectric elbow-wrist-hand orthosis (EWHO) currently on the market for civilians as well as numerous VA facilities across the United States. Surface sensors - built into the orthosis and located over the upper arm and forearm muscles - detect the user's electromyography (EMG) signal once he/she initiates a muscle contraction. The EMG signal activates the motors on the orthosis to move the elbow or hand in the desired direction, proportional to muscle output.

CASE DESCRIPTION

The participant is a 62 yr. old veteran, who served in the Marines as a Corporal and in the Navy as a Seaman and is a recipient of a Purple Heart. He is a right hand dominant male who presented with dysarthria and left hemiplegia on 12/2/2013. Past medical history is positive for hypertension, tobacco use, a previous transient ischemic attack in 2010 and hyperlipidaemia. The participant began traditional outpatient occupational therapy (OT) in January 2014 to treat the functional deficits associated with his left hemiparesis, and was prescribed a custom myoelectric EWHO (without grasp capability) in April 2014. The patient met the criteria for the myoelectric EWHO including intact cognition, no UE contractures and sufficient EMG signal strength in his bicep and triceps to power the orthosis. Measurements and a cast were taken of his affected arm and a custom myoelectric EWHO was manufactured. The participant continued to participate in OT with his orthosis until August 2014, at which point he was re-evaluated for the advanced model that includes additional myoelectric grasp capabilities. The participant received his upgraded custom myoelectric EWHO in November 2014. At this time, the overall fit and comfort of the device was assessed, appropriate sensor locations were found and the device was programmed with the appropriate level of assistance using the manufacturers programming software. At the time of starting to use the myoelectric EWHO with grasp, the participant presented with unresolved UE deficits with range of motion, strength, fine and gross motor skills and functional use of the paretic left arm.

METHODS

The participant completed a total of 21, 1 hour outpatient OT sessions that began with traditional OT interventions for 3 months, such as functional electrical stimulation, mirror therapy, dynataping, massage wand, fluidotherapy, proprioceptive neuromuscular facilitation, therapeutic exercise, active-assisted and passive range of motion and task oriented/occupation based interventions, and then incorporated the custom myoelectric EWHO without grasp after it was delivered. The participant received training in proficient use of the orthosis: donning/doffing technique, repetitive task practice drills and application of the orthosis during multi-step functional tasks. He was also given a home activity plan and a wearing schedule, beginning at 30mins daily, in order to build up endurance and facilitate functional use of his affected extremity. After upgrading to the custom myoelectric EWHO with grasp, the participant completed an additional 14 sessions under the supervision of his therapist and fitting prosthetist, dedicated to mastering the operation of the myoelectric EWHO with grasp and utilizing it during daily functional tasks. A functional training protocol from the myoelectric EWHO manufacturer was incorporated at this time. Outcome testing throughout treatment included ROM (range of motion) testing, strength (Manual Muscle Testing), MAS (Modified Ashworth Scale) to assess spasticity and functional task assessment/observation. Once the participant upgraded to the myoelectric EWHO with grasp, the Fugl Meyer assessment was also utilized to quantify progress.

RESULTS

Active left upper extremity ROM and strength both increased significantly (Table1 ROM, Table 2 Strength). He also demonstrated an improved ability to incorporate his affected extremity (while wearing his orthosis) into a wide variety of bilateral, gross motor ADLs and IADLs such as carrying a laundry basket (Photo 1), lifting heavy objects (e.g. a chair), using a tape measure, meal preparation and opening doors.

Photo 1

These results demonstrate how working with a custom myoelectric EWHO resulted in remediation of UE deficits, in particular wrist and hand function. Functionally, the participant demonstrated substantial improvements in daily use of his paretic left arm, and an increase in his overall level of independence and function at home and in his community. While wearing the myoelectric EWHO, the participant was able to use his left arm to carry weighted objects bilaterally, to stabilize objects such as a cup or plate and to put items away in overhead cabinets. He also demonstrated independence with household chores such as laundry, meal preparation and light cleaning tasks. Information from the manufacturers' survey and home log (September 2015 – March 2016) show that the participant was able to complete tasks such as sweeping/mopping the floor, washing and folding clothes and moving/lifting chairs. During this 6 month timeframe, the participant logged 25 entries and wore his myoelectric EWHO at home for a total of 34 hours during that time. The participant wore his myoelectric EWHO between 30 mins – 2 hours each time and reported an average satisfaction rating of 8.78/10 (0 = not satisfied at all, 10 = extremely satisfied). Overall, the participant reported a high level of satisfaction, improvement in his quality of life and increased functionality of his paretic arm. He also reported a few areas for improvement. These included addressing technical glitches with the sensors and electronics, decreasing the weight and bulk of the myoelectric EWHO (4lbs); redesigning the harness, increasing the battery life and making the myoelectric EWHO waterproof so he could wear in wet conditions. Battery life and fatigue were the most common reasons for needing to stop use of the myoelectric EWHO. The participant was able to self-don his myoelectric EWHO independently in 3-5 minutes, but he did express the need to practice and that it was difficult when he first tried to do it alone. Interestingly, the participant noted that if he stops using his myoelectric EWHO for longer than 2 days, his arm – in particular his hand and fingers – start to stiffen, loose ROM and function.

Table 1 ROM

	May 2014 (MyoPro Classic)	Sept 2014	Nov 2014 (MyoPro Motion G)	Jan 2015	March 2015	May 2015	March 2016
AROM (degrees)							
Shoulder flex	91 (supine)	133 (supine)	120 (seated with compensation)	120 (seated with comp)	155	WFL	WFL
Shoulder abd/add	90 (supine)	105 (supine)	82 (seated with compensation)	82	120/ Full	WFL	WFL
Elbow flex	97 (seated)	WFL (supine)	125 (seated)	125	130	WFL	WFL
Elbow ext	-30 (seated)	WFL (supine)	-20 (seated)	-20	0	WFL	WFL
Supination/ Pronation	NT	NT	60/40	NT	60/full	NT	NT
Wrist flex	20 with gravity assist	20	20	35	40	55	NT
Wrist ext	40 with compensation	45	50	60	65	70	NT
Radial/ ulnar deviation	NT	NT	NT	NT	NT	15/30	NT

Table 2 Strength

	May 2014 (MyoPro Classic)	Sept 2014	Nov 2014 (MyoPro Motion G)	Jan 2015	March 2015	March 2016
Strength (MMT w/o MyoPro)						
Shoulder	3-/5	3-/5	3-/5	3-/5	3-/5	4/5
Elbow	3-/5	3-/5	3-/5	3-/5	3/5 Pro/Sup 3/5	4/5
Wrist	3-/5	3-/5	3-/5	3-/5	Flex 3-/5 Ext 3/5	NT
Hand	3-/5	3-/5	3-/5	3-/5	2-/5	NT
Grip (L hand)	21 lbs	31 lbs	36 lbs	40 lbs	47 lbs	NT
Lateral pinch	NT	NT	NT	NT	11 lbs	NT

DISCUSSION

Regular and consistent use of the custom myoelectric EWHO in conjunction with traditional OT has resulted in many benefits and positive functional outcomes for the participant. The myoelectric EWHO was shown to provide both assistive device benefits as well as rehabilitative benefits. While wearing his orthosis, he is able to be more independent with daily ADLs and IADLs such as meal preparation, folding and washing clothes, accessing items in overhead cabinets, bilateral lifting tasks (e.g. laundry basket, dining chair), sweeping/mopping and stabilizing objects such as cups and plates. Once fitted with the myoelectric EWHO with grasp, the participant demonstrated significant improvement in his affected hand function in particular, again both without and without the myoelectric

EWHO donned. Consistent use of this myoelectric orthosis over time has also resulted in dramatic therapeutic improvements in the range of motion and strength and a reduction of spasticity in his paretic arm. The participant's comment that not using his myoelectric EWHO with grasp for longer than 2 days results in his arm and hand stiffening and losing function, suggests that the changes he has experienced in his arm are largely due to the integration of the myoelectric EWHO into his therapy regime (Table 3 Spasticity). Looking at the data from November 2014 onwards (in particular March 2015 – October 2015), we see improvements in gross grasp as well as lateral pinch strength and wrist ROM and blossoming ulnar and radial deviation. With this new found hand function and increased UE strength overall, the participant demonstrates competence with tasks such as opening doors and cupboards with his affected hand, as well as using a measuring tape (bilateral task), picking a phone off the hook and holding papers, all without using the myoelectric EWHO. Results from the Fugl-Meyer (Table 4 Fugl Meyer) assessment (specifically in the wrist, hand and coordination/speed categories) also support a significant training effect and remediation of upper extremity paresis. Since the participant also engaged in traditional OT, it should be noted that it is challenging to separate the progress made by just traditional therapy versus the effects of the myoelectric EWHO without grasp capabilities. It is reasonable to conclude that incorporating a myoelectric EWHO into a comprehensive treatment program offers excellent results. However, given the participant's level of hand function immediately prior to receiving the myoelectric EWHO with grasp, it does seem reasonable to conclude that the added myoelectric grasp feature contributed directly to the participant's significant distal fine and gross motor recovery.

Table 3 Spasticity

	Jan 2014	May 2014 (MyoPro Classic)	Sept 2014	Nov 2014 (MyoPro Motion G)	Jan 2015	March 2015	March 2016
Spasticity (Modified Ashworth Scale)							
Elbow Flex	1+	1	0	0	0	NT	0
Elbow Ext	1+	0	0	0	0	NT	0
Wrist Flex	1+	0	0	0	0	NT	0
Wrist Ext	1+ with 2 beats clonus	1	1	0	1 with clonus	NT	0

Table 4 Fugl Meyer

	March 2015	March 2016
Upper Extremity	31/36	31/36
Wrist	9/10	10/10
Hand	9/14	14/14
Coordination/Speed	3/6	6/6
Sensation	11/12	11/12
Passive Joint Motion	24/24	24/24
Joint Pain	24/24	24/24
Total Score	111/126	120/126

CONCLUSION

This custom myoelectric EWHO was shown to provide this veteran with an increased ability to move and use his affected arm in a variety of functional tasks, in particular bilateral tasks. The participant wears his myoelectric EWHO on and off throughout a day and reports improvements in his independence with daily functional tasks and overall quality of life. This participant has also demonstrated significant recovery in his affected upper extremity including improved active range of motion at the shoulder, elbow and hand, improved strength and a reduction in tone. This case report highlights both short term assistive and functional benefits as well as long term rehabilitative benefits of a myoelectric orthosis. These devices offer an exciting opportunity for other individuals diagnosed with chronic stroke or brain injury to make advancements towards their recovery and independence, and warrant additional research into the application of custom myoelectric EWHOs for veterans and active duty personnel.

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DEVELOPMENT OF A WIRELESS MULTICHANNEL MYOELECTRIC IMPLANT FOR PROSTHESIS CONTROL

Daniel McDonnall, Scott Hiatt, Brian Crofts, Christopher Smith, Daniel Merrill

Ripple LLC, Salt Lake City, UT, USA

INTRODUCTION

Complete functional adoption of upper limb prostheses is unacceptably low. Myoelectric device rejection rates are comparable to those of body-powered prostheses, even though these devices should be capable of providing better function. Amputees cite awkward use and lack of perceived utility of their myoelectric prostheses, as well as dissatisfaction with the ability to perform ADLs. Ultimately, poor control of myoelectric systems limits the adoption of advanced hand prostheses.

Prostheses manufacturers have released substantially improved prosthetic arm technology in the last decade; however, a well-documented challenge with the implementation of current myoelectric devices is the common use of only two surface EMG electrodes for collection of control signals. Limitations in the control signals extracted from surface EMG signals prevent the implementation of advanced control algorithms and intuitive movement. As a result, these advances prostheses still require serial selection and control of individual joints and grips resulting in slow, unnatural motions.

SYSTEM DESIGN

Ripple has developed an implantable system which simultaneously records 32 channels of myoelectric data from multiple residual muscles, and transmits these data to an external transceiver placed in the prosthetic socket. Our objective is to provide simultaneous multi-degree of freedom prosthesis control, ultimately providing an intuitive control experience. This approach supports a high number of independent control signals and provides access to EMG from deep muscles that cannot be accessed with surface electrodes.

The system comprises a hermetic implanted module from which nine EMG leads emerge. Eight of the leads contain four electrode sites each for 32 total recording channels. A ninth lead provides the reference electrode. The implant receives power inductively from an external transceiver and sends digitized EMG data to the external transceiver via infrared light. By using a single subcutaneous module for telemetry from which several leads emerge, power coupling efficiency remains high.

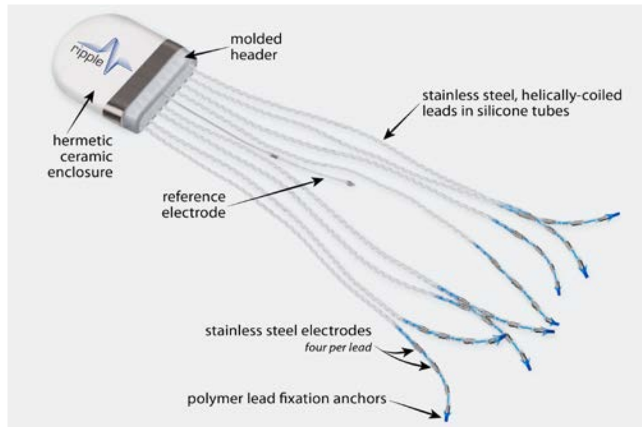

Figure 1. Wireless 32-channel Myoelectric Implant

RESULTS

We have demonstrated high data rate transmission using infrared light in chronically implanted canines. Devices were implanted in deltoideus and the long and lateral heads of triceps. Recorded EMG demonstrate very low noise and clearly indicate antagonistic activity of the gait muscles. Recordings are stable over the 6-month implant period.

We have completed and pass safety, electromagnetic compatibility, biocompatibility, sterilization, hermeticity, impact, lead flexion, and performance testing.

CONCLUSIONS

These efforts demonstrate the ability to amplify and transmit muscle signals and confirm safety and performance requirements. This approach has the potential to provide simultaneous multi-degree of freedom prosthesis control, especially if used with dexterous prostheses, surgical reinnervation techniques (TMR and RPNI), and advanced algorithms.

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