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CONSIDERATIONS FOR PROSTHETIC MANAGEMENT OF ELECTRIVE UPPER LIMB AMPUTATION FOLLOWING BRACHIAL PLEXOPATHY INJURY WITH AN ILLUSTRATIVE CASE STUDY

Branden Petersen CP LP[1], Phillip M. Stevens Med, CPO[1,2]

[1]*Upper Limb Prosthetics Program, Hanger Clinic, San Antonio TX, USA*

[2]*Department of Clinical and Scientific Affairs, Hanger Clinic, Austin TX, USA*

ABSTRACT

Most major upper limb amputations result from trauma. Occasionally, these traumatic injuries include localized inury to the nerves of the brachial plexus. Patients may seek elective amputation following severe brachial plexus injury (BPI) [1]. The evaluation and development of a prosthetic treatment plan for this cohort often involves surgical considerations prior to prosthetic intervention. This paper will review the types of injuries that can be sustained to the brachial plexus nerve complex as well as surgical options associated with brachial plexopathy cases. A representative case study will document the surgical and prosthetic considerations of an individual that was involved in a motor vehicle accident that left him with a flail upper limb secondary to BPI. For this case presentation long term follow-up, patient perceptions and functionality will be discussed. .

INTRODUCTION

Brachial plexus nerve injuries can have devastating consequences to an individual's overall functionality and quality of life [2]. These significant injuries can lead to the inability to return to premorbid occupations and activities. The deficits associated with BPI may be partial or full and can often require months to years to fully realize the full possibilities of functional return [3,4]. This realization of requires the consideration of critical surgical timeframes which are often unknown or neglected, undermining long term outcomes for those with BPI cohort [5].

Patients may live with their flail limbs for years, at times supported and protected in various bracing systems. Over time, gravitational forces acting on the neurologically impaired shoulder muscles and glenohumeral joint may cause the limb to sublux. In such cases the supporting musculature and ligaments are no longer sufficient to maintain the humeral head in the glenoid fossa. In addition, without protective sensation, this cohort can sustain severe injury to their limb without their immediate awareness. In many cases discussions of elective amputations are driven by continued inadvertent injury to the flail and insensate hand and limb. Elective amputations should not be considered a failure but an opportunity for reconstruction [6,7]. Collective effort from the patient, patient's family, surgeon, rehabilitation physician, prosthetist, occupational and physical therapists will be key in developing the best rehabilitation plan.

BPI TYPE AND SEVERITY

The type and severity of a BPI are a function of the mechanism, extent and location of the injury. *Nerve root avulsion* injuries occur when the nerve root is torn from the spinal cord and cannot be surgically repaired. As the name implies, a *nerve stretch injury* results from a mild stretch of the nerve that may allow some functional return over time. In such injuries it is generally accepted to receive occupational/ physical therapy and allow time for functional return. *Nerve rupture* represent a more forceful nerve stretch injury that may result in partial or full nerve tears. Such ruptures may be repaired surgically depending on the location of the injury.

Depending on the mechanism of injury, various portions of the brachial plexus can result in different palsy presentations. These include upper trunk, lower trunk, and pan nerve injuries, each with specific clinical presentations. *Upper trunk palsy* of the brachial plexus is often the result of the arm being pulled down while the head is forcefully pushed to the opposite side of the arm involved [8]. Such injuries generally results in muscle weakness around the shoulder joint as well as elbow positioning capabilities, with compromise to the deltoid, rotator cuff, and biceps musculature. *Lower trunk palsy* can result from injuries where the arm is forcefully pulled upward. These injuries will generally result in functional loss at the hand of the affected extremity, with claw-like hand deformities commonly occurring. *Pan palsy* is when both upper and lower trunks are injured resulting in complete paralysis of the musculature around the shoulder, elbow and hand. This is often referred to as flail limb.

BPI TREATMENT APPROACHES

There are several treatment approaches that can be considered when the nerves of the brachial plexus nerve are injured. Viable options depend on the where the nerves were injured as well as the extent of the associated nerve damage. They include nerve repairs, nerve grafts, nerve transfers, tendon and muscle transfers, and joint arthrodeses.

Nerve repair can be done to surgically restore the cut ends of nerves. These can assist in stabilizing joints, restoring elbow functionality and a sensible hand following nerve injury [9]. *Nerve grafting* occurs when a healthy nerve from another part of the body is used to replace a missing or damaged nerve. *Nerve transfers* from one muscle to another can occur to provide alternate innervation to a major muscle group when the primary innervation has been injured. *Tendon and muscle transfers* can be performed to address significant functional deficits by restoring key joint movements.

When surgical reconstructive efforts fail to yield a functional hand or elbow, some patients may wish to pursue elective amputation of the flail limb. This is often coupled with glenohumeral arthrodesis, and is performed when there is adequate muscle strength in the trapezius, levator scapulae, rhomboids, and serratus anterior [10,11]. The generally accepted position of the glenohumeral joint is in 30 degrees flexion, 30 degrees abduction, and 30 degrees internal rotation [12,13]. In general, a 4 month postoperative period is required for fusion occur [13].

While there are many different references to these fusion angles discussed the literature, the guiding principles are to pace the residual limb in enough glenohumeral joint abduction to clear the axilla as well as allow the patients to perform axillary hygiene, to place the residual limb in enough forward glenohumeral flexion to bring the arm and terminal device of the prosthesis toward the midline for functional activities and minimize subluxation of the glenohumeral joint

CASE STUDY

Written informed consent was obtained from the patient prior to his inclusion in this paper. In 2012 our case study of a 22 year old male was involved in a snowmobiling accident that left him with nerve avulsion injuries to his the brachial plexus resulting in a flail limb. This individual worked on a family dairy farm and expressed an interest in returning to his family business. He described himself as a "hands on" individual desiring to return to as much functionality as possible. The patient's contralateral scapular range of motion was with in normal range, and contralateral scapular strength was sufficiently strong to operate cable operated components. The medical team discussed several options with the patient, ultimately choosing shoulder joint arthrodesis coupled with an elective elbow disarticulation as the best option to restore functionality for his lifestyle (Figure 1 and 2).

Figure 1: Initial limb following should fusion in 2012

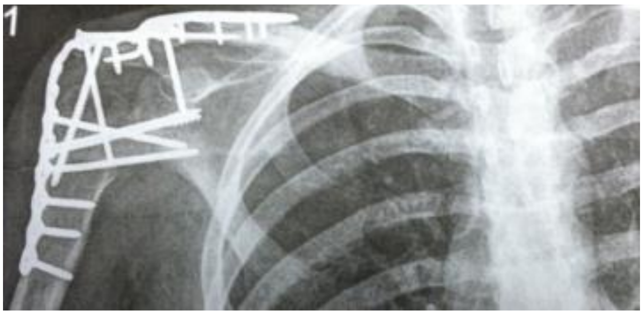

Figure 2: Internal hardware in initial shoulder fusion

Following the elective surgical procedures and prosthetic fitting the patient expressed satisfaction with his ability to move his arm again (Figure 3). He was able to demonstrate full functionality of the prosthesis in both elbow control and terminal device function. He reported regular use of his cable operated device on his family farm daily running equipment, carrying and manipulating objects. Regular clinic visits for frequent repairs to his device support the patient's reports of sustained regular use of the device for heavy duty activities on his farm.

Figure 3: Initial body powered prosthesis with work hook, external locking elbow joints and chest strap.

At 8 years post injury, our case demonstrated several anatomic characteristics common to sustained severe BPI. (Figure 4). For significant nerve root avulsion injuries these include atrophy of the deltoids, infraspinatus, supraspinatus, biceps, and triceps muscles. Bony anatomy becomes very prominent, including the spine of the scapula, acromion, and coracoid process. The lack of protective sensation, diminished muscular padding, and significant prominence of bony anatomy creates significant design consideration when designing a prosthetic socket for individuals with brachial plexopathies.

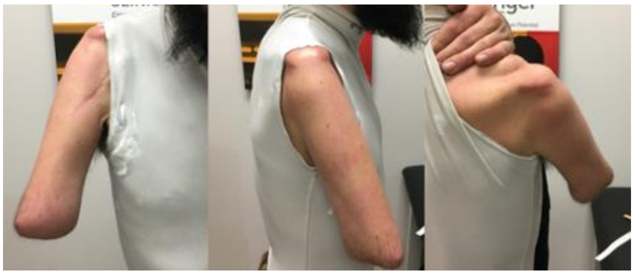

Figure 4: Limb presentation 8 years post injury, elective amputation and shoulder arthrodesis characterized by soft tissue atrophy and significant bony prominence around the shoulder.

In 2018 the internal fixation hardware from the arthrodesis was removed from the patient's limb due to harness pressures and prosthesis usage on his highly atrophied limb (Figure 5). The surgeon evaluated and determined that the glenohumeral joint had fused well enough to remove most of the internal hardware and screws.

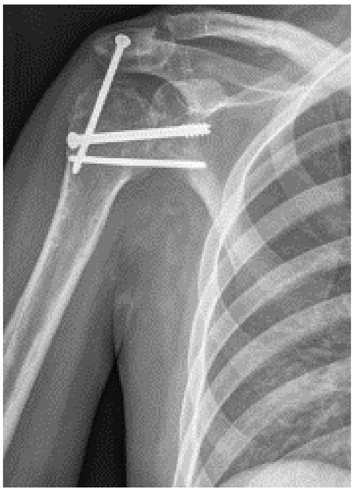

Figure 5: Subsequent removal of most internal fixation hardware with adequate bony fusion

In 2020 the patient continues to work on his family farm as well as running his own business offering handy man services. He is currently married and has children. He continues to wear his prosthesis every day full time for all of his home and work activities (Figure 6). He continues to need repairs to his prosthesis indicating that indeed he uses the device daily and in a heavy duty capacity.

Figure 6: Current body-powered prosthesis

CONCLUSION

The prosthetic management of individuals with brachial plexopathies can be challenging and should involve several medical professional to develop the best treatment plan with optimum outcomes. Brachial plexus surgical interventions can improve the overall functionality when considering prosthetic intervention. In this particular case study the shoulder arthrodesis produced a very functional outcome for almost a decade. The patient actively utilizes his limb and prosthesis for most of his activities.

This case study does not reflect every patient's particular situation. This patient is a young, active male that has excellent scapular strength and range of motion. In some BPI cases patients may not be able to generate the required force and excursion requirements to operate a body powered systems and require externally powered components to create the desired functionality (14). Patients will require individual evaluation to determine their functional capabilities following BPI so that an appropriate prosthetic treatment plan can be created.

These cases present many challenges to the rehabilitation team. Decisive surgical decision making can create a limb that is better reconstructed for improved prosthetic functionality.

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DIFFERENCES IN MULTIGRIP MYOELECTRIC HANDS FOR FACILITATING ACTIVITIES OF DAILY LIVING

Andreas Kannenberg*, Russ Lundstrom*, Arri Morris*, Shawn Swanson Johnson**

*Otto Bock Healthcare LP, Austin, TX, USA

**SSJ Rehab Services LLP, Houston, TX, USA

ABSTRACT

Most transradial amputees are fitted with a prosthetic hand but use it actively for only 50% of activities of daily living (ADLs). Studies with the multigrip Michelangelo hand reported that many patients perceived ADLs easier to perform than with a conventional prosthetic hand [3] and could also demonstrate improvements in objective ADL performance. Other multigrip hands available on the market offer more grip types than the Michelangelo hand but have not yet been subjected to published clinical studies. Thus, it is unknown whether more grip types result in even greater perceived ease of ADLs execution.

Subjects wearing the bebionic or i-limb hands were assessed with the same hybrid questionnaire as used in the previous Michelangelo study. Demographic information on all subjects was also collected. The results were then compared to the historical data collected in the previous Michelangelo study.

Data were available from 36 unilateral subjects with transradial amputations, 10 each wearing a bebionic or i-limb, respectively, and 16 historical datasets of subjects who used a Michelangelo and conventional hand, respectively.

Means for ease scores and "useful" ratings across 23 ADLs did not differ between the multigrip hands but were better than those for the conventional hands. There were no statistical differences between the 3 multigrip hands. The mean numbers of ADLs by usefulness and method of use (prosthesis actively used to grasp, prosthesis passively used to stabilize, assistance of residual limb, sound hand alone) rating were also similar.

Analyzing the ease of individual activities, Michelangelo mean ease scores for several activities showed modest positive differences compared to conventional myoelectric hands. In contrast, the bebionic profile indicates fewer activities that were scored easier than conventional myoelectric compared with Michelangelo profile, but the difference in the scores for several activities were much greater than for the Michelangelo hand. For the i-limb, there were also several activities for which differences in the mean scores compared to conventional myoelectrics were much greater than that for Michelangelo.

In conclusion, all multigrip myoelectric hands may reduce the difficulty for performing ADLs vs. conventional hands. However, the availability of more grip types in a hand does not necessarily result in greater ease of performance of ADLs in general. Interestingly, the 3 multigrip hands studied showed different activity profiles that they facilitate. For some activities, there was a clear advantage for some hands over others. Thus, clinicians' knowledge of the patients' functional needs and the differential features of the available multigrip hands is crucial for selecting the best suitable hand for an individual patient. In addition, this study also highlights the need for more sophisticated control (e.g. pattern recognition) that facilitates easier and more intuitive access to a greater number of grips in a prosthetic hand than the current 2-channel myoelectric control.

INTRODUCTION

Multiarticulating and multigrip myoelectric prosthetic hands have been available on the market for about 15 years now. A study published in 2015 (3) demonstrated improved ease of performing activities of daily living (ADL), increased usefulness, and more active use to grasp objects with the Michelangelo® hand (Ottobock, Germany) that offers 7 grip types and hand positions as compared to standard myoelectric hands that offer only the opposition grip. The purpose of this study was to gather information on the perceived ease, usefulness und way of use of two multiarticulating hands with multiple grip options, i-limb (Össur hf, Iceland; 12-18 grip types and hand positions, depending on version) and bebionic (Ottobock, Germany; 14 grip types and hand positions), and to compare the results with those previously published for Michelangelo hand [3] to answer the question whether or not more available grip types result in more perceived functionality of a prosthetic hand.

METHODS

IRB approval was obtained for prospective data collection. The Multigrip Myoelectric Hand Survey was launched in March of 2016. Data collection started in June of 2016 and was completed in September of 2017. All subjects were asked to complete two questionnaires. The first was a combination (Pröbsting et al, 2015) of a modified Orthotics and Prosthetics User Survey – Upper extremity Functional Status (OPUS-UEFS) [5, 6] and the Prosthetic Upper Extremity Functional Index (PUFI) [7]. The modified OPUS-UEFS asks subjects to rate how easily he/she can perform ADLs with the prosthetic hand, and the addition of the PUFI asks about how each ADL was performed and how useful the prosthesis was for each ADL. The second questionnaire was a set of questions including the reasons for selecting the type of hand, the most frequently used grip patterns, and ranking the importance of hand features. Demographic information on all subjects was also collected. This included age, sex, years of prosthetic use, amputation side and etiology of the amputation. These results were then compared to previous data collected on the Michelangelo hand [3].

RESULTS

Patient Population

Data were collected from 25 subjects using either a bebionic or i-limb hand. Five subjects were excluded from the final analysis; two had above-elbow amputations and the other three were bilateral users. 70% of these subjects were male, and 30% were female. The results from 20 i-limb and bebionic users with unilateral transradial amputations were then compared to results from a previous study of 16 male myoelectric hand users fitted with a Michelangelo hand. The mean age for the i-limb group was 50.4±17.6 years, while the mean age for the bebionic group was 37.4±14.2 years. In comparison, the mean age for the Michelangelo group was 43.9±17.3 years. Bebionic users had had their device for an average of 1.65±1.10 years, while i-limb users had had their device for an average of 2.08±1.87 years. In contrast, the Michelangelo users had only been using their myoelectric hand for 0.24±0.18 years.

Clinical Results

The means for ease scores across the subset of 23 ADLs for each of the multi-grip myoelectric hands were remarkably similar (Table 1), but all higher than the scores reported for the conventional myoelectric hands in the study with the Michelangelo hand [3].

Table 1: Ease scores and # activities for which hand was rated Very Useful or Useful for 23 ADLs

Mean ± SD	**Conventional**	**Michelangelo**	**bebionic**	**i-limb**
Ease score for performing 23 ADLs	27 ± 9.7	37 ± 12.7	33 ± 13.5	35 ± 14.9
# activities for which hand was rated Useful	15.7 ± 3.6	17.9 ± 4.0	17.2 ± 4.9	17.7 ± 4.9

The mean numbers of ADLs by usefulness rating were also similar and, likewise, higher than the mean for the conventional myoelectrics (Table 2).

Table 2: PUFI Prosthesis Usefulness Ratings by Prosthetic Hand

Mean #Activities ± SD	**Conventional**	**Multi-grip Myoelectric Hands**		
		Michelangelo	***bebionic***	***i-limb***
Not Useful	11.7 ± 3.1	9.8 ± 3.9	9.7 ± 4.0	8.9 ± 5.0
Useful	4.9 ± 3.4	4.2 ± 2.7	3.8 ± 3.1	6.1 ± 2.6
Very Useful	6.4 ± 4.1	9.1 ± 4.3	9.2 ± 3.7	7.2 ± 4.4

The mean number of ADLs by way of use was also similar, with the number of ADLs performed by using both hands and the prosthesis *actively* was slightly higher for bebionic, and the number using only the sound hand slightly lower for bebionic (Table 3).

Table 3: PUFI Method Assessment by Prosthetic Hand

Mean #Activities ± SD	**Conventional**	**Multi-grip Myoelectric Hands**		
		Michelangelo	***bebionic***	***i-limb***
Both hands, prosthesis actively	7.1 ± 4.1	9.3 ± 4.6	10.7 ± 2.9	9.8 ± 3.0
Both hands, prosthesis passively	2.4 ± 2.4	1.8 ± 1.9	2.1 ± 2.5	2.2 ± 1.8

The comparative ease of performing the 23 ADLs of the OPUS-UEFS with the multigrip or conventional prosthetic hands showed that each of the advanced hands had strengths and weaknesses (Figure 2). While the Michelangelo hand scored somewhat better than the conventional hands across the board (except 2 ADLs), the bebionic and iLimb hands scored considerably better in some (6 or 9, respectively) but also much worse than the conventional hands in some other (5 or 3, respectively) ADLs.

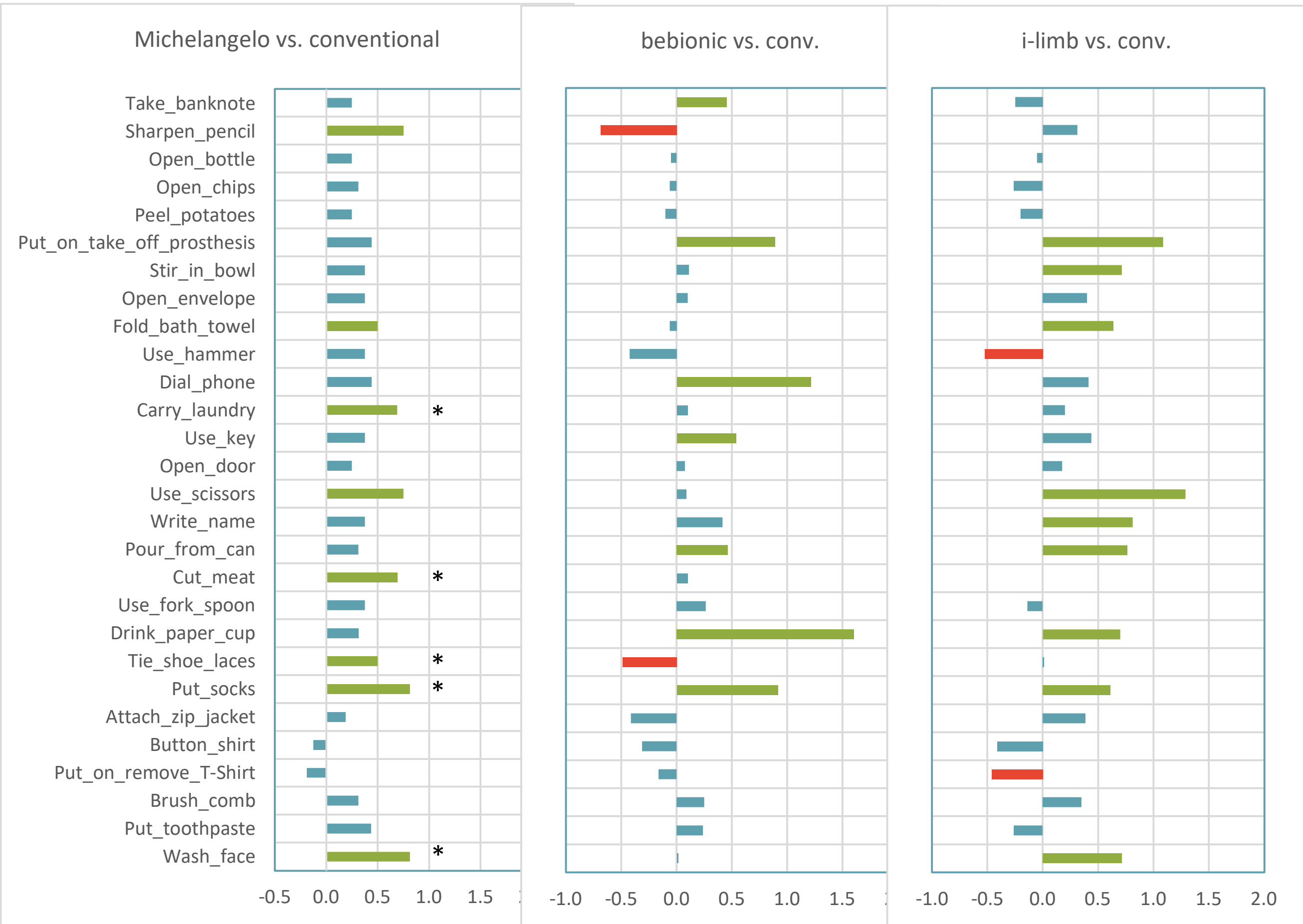

Figure 2 ADL Ease Profiles. Differences in mean ease scores by ADL for multi-grip hands compared to conventional myoelectric hands. **Red** bars signify decreased ease, **Green**, increased ease approaching a clinically meaningful difference, and **Blue** differences less than what could be considered clinically meaningful. * p<0.05 as reported in the Michelangelo study.

DISCUSSION

The aim of the study was to investigate whether more than 12 grip types and hand positions offered by a myoelectric hand might further reduce the difficulty of ADLs as shown for the Michelangelo hand with its 7 grips and hand positions.

Overall, the ease, usefulness and way of use of all three multigrip hands did not significantly differ compared to each other. Compared to conventional myoelectric hands, there was an overall improvement in ease and usefulness ratings and an increase in ADLs in which the multigrip hands were actively used to grasp. While Michelangelo showed moderate improvement in all but two ADLs, bebionic and i-limb showed considerable improvement for some ADLs but also substantial decline in ease and usefulness for some other ADLs. This suggests that there is no "perfect" posthetic hand and that clinicians must match the functional ADL needs of each patient with the hand that meets these specific needs best.

CONCLUSIONS

All multigrip myoelectric hands may reduce the difficulty for performing ADLs vs. conventional hands. However, the availability of more grip types in a hand does not necessarily result in greater ease of performance of ADLs and greater perceived usefulness in general. Interestingly, the 3 multigrip hands studied showed different activity profiles that they facilitate. For some activities, there was a clear advantage for some hands over others. Thus, clinicians' knowledge of the patients' functional needs and the differential features of all multigrip hands available on the market is crucial for selecting the best suitable hand for an individual patient. In addition, this study also highlights the need for more sophisticated control (e.g. pattern recognition) that facilitates easier and more intuitive access to a greater number of grips in a prosthetic hand than the current 2-channel myoelectric control.

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Innovative Outcome Measurement in Upper Limb Prosthetic Rehabilitation
Authors: Kerstin Baun MPH OTR/L, John Miguelez CP, FAAOP(D), Dan Conyers CPO, FAAOP

Introduction: Outcome measure development has long been recognized as a need in the field of upper limb prosthetic rehabilitation to map individual patient progress, highlight needs for component development and cost justification.[1] In response, several measures have been created and proven valid.[2-10] However, gaps remain in the effort to capture the complex facets of prosthesis use that ultimately determine success—physical, psychological, social and environmental. This paper describes a suite of 3 measures developed over the past decade that together capture these complex facets more completely. These measures include the Capacity Assessment of Prosthesis Performance of the Upper Limb (CAPPFUL), the Comprehensive Arm Prosthesis and Rehabilitation Outcomes Questionnaire (CAPROQ) and the Wellness Inventory (WI). Each measure will be described individually, including validation data and their value and potential for guiding patient care, device selection and development and cost justification. All studies related to measure development were approved by the WIRB.

Outcome Measure Descriptions:
Capacity Assessment of Prosthesis Performance of the Upper Limb (CAPPFUL):
CAPPFUL is designed as a versatile, low-burden measure of prosthesis performance for any UL functional prosthetic device type and any UL amputation level. Unlike most measures of performance, CAPPFUL assesses overall performance and 5 functional performance domains during completion of 11 tasks. These require movement in all planes while manipulating everyday objects requiring multiple grasp patterns. Performance domains include control skill, adaptive and maladaptive compensatory movement, component utilization and time for task completion. Performance is scored relative to function of a sound upper limb, preventing ceiling effect. For the individual patient, scores within performance domains can target further training needs and assist the treatment team in focusing on optimal strategies to develop performance and function. Multiple administrations assist the team in objectively measuring improvement in performance with the prosthesis over time. Information gathered assists not only to guide therapeutic training but also to determine need for components and/or fit and design modifications. Cumulative data, across prosthetic options and levels of amputation, can establish expectations for current devices, provide reimbursement justification and set goals for future product development. Current administrations including validation study subjects exceeds 200.
Validation: Psychometric evaluation indicates good interrater reliability, internal consistency, known-group validity, and convergent and discriminant validity. Specifically, interrater reliability was excellent for scoring on the task, domain, and full-scale scores (intraclass correlation coefficientsZ.88-.99).Internal consistency was good (aZ.79-.82). CAPPFUL demonstrated strong correlations with measures of hand dexterity or functioning (rsZ.58 to .72) and moderate correlation with self-reported disability (rZ.35).[11]

Comprehensive Arm Prosthesis and Rehabilitation Outcomes Questionnaire (CAPROQ):
The CAPROQ is designed to measure patient reported outcomes in key facets of rehabilitation for adults with UL absence or loss: perceived function, satisfaction and pain. It is a low burden measure to guide individual patient care, as well as assess and improve care models and inform future prosthesis selection and development for the UL loss community. Results inform the treatment team of current status and change of status through the continuum of care and assists with targeting of further training needs as well as providing valuable feedback regarding prosthesis fit and function. CAPROQ cumulative data, across prosthetic options and levels of amputation, provides patient perspectives regarding currently available devices, potential reimbursement justification and guidance for future product development. Original CAPROQ was administered 687 times and since validation study completion, over 100 administrations have been completed with more being added weekly.
Validation Study: Psychometric evaluation with 261 subjects demonstrated adequate-to-strong factor loading on each subscale, good-to-excellent internal consistencies for measure subscales and moderate-to-strong convergent validity. Specifically, confirmatory factor analysis indicated adequate-to-strong factor loading on each subscale: satisfaction (.623-.913), perceived function (.572-.860) and pain (.422-.834). Internal consistencies for the measure subscales were good-to-excellent (.89-.95) and convergent validity indicated moderate-to-strong statistically significant associations between the CAPROQ and the measures tested—Disabilities of Arm Shoulder and Hand questionnaire (DASH), Trinity Amputation and

Prosthesis Experience Scale Revised (TAPES-R) and Brief Pain Inventory (BPI). Currently this validation study is in submission process for peer review.

Wellness Inventory:
The wellness inventory screen was designed to inform prospective prosthesis recipients of how they compare to other people in areas such as coping style, perceived quality of life and other areas that have been shown in the rehabilitation research literature to have an impact on how people perform after acquiring a physical disability. It is a short battery of seven validated screening instruments that measures resilience[12], health-related quality of life (OPUS)[13], pain (SF-36/12)[14], depression[15], alcohol use (AUDIT-C)[16], drug use/misuse, and posttraumatic anxiety (PC-PTSD)[17]. In 2014, analysis of results from 123 patients was conducted confirming high prevalence of mental health concerns in this sample. The WI seeks to promote patient self-understanding during treatment and beyond and, if indicated, to mobilize provision of mental health services by appropriate providers. Re-administration of the WI 6-12 months post prosthesis fitting can determine change in status through the continuum of care. Since inception, over 500 WIs have been administered across seven centers in the US.

The WHO International classification of function[18], identifies 3 domains (Body Functions/Structures, Activities, Participation) and 2 contextual factors (environmental and personal) in complex relationship with a health condition such as upper limb difference. The CAPPFUL addresses Body Structures and Function, and Activities through performance assessment. The CAPROQ, through patient report, addresses all three domains along with environmental factors. The WI, through structured interview, covers personal factors. However, not all of these assessments are appropriate for administration at all times in the continuum of care. The Wellness Inventory is most aptly used early in the rehabilitation process and can assist the patient in decision making regarding whether to pursue psychological care and provides insight for the treatment team in terms of factors that might impact rehabilitation. The WI can be re-administered subsequently to determine change in status or identify further needs. The CAPROQ can also be administered pre prosthesis fitting to obtain baseline data in areas of pain and perceived function. Re-administration post prosthesis fitting tracks changes in these areas as well as capturing satisfaction data. The CAPPFUL is strictly designed for post prosthesis fitting use; with initial administrations, training needs can be identified as well as potential design and component modifications needed. Subsequent administrations can demonstrate progress and further training opportunities, component/prosthesis effectiveness and overall return of function.

Conclusion: Goals for outcome measures vary from ensuring provision of excellent individual patient care to assessment of currently available devices to justification of cost related to both current of future products and more. When administered in concert, the measures described (CAPPFUL, CAPROQ and Wellness Inventory) provide complimentary data relevant to each stage of care and capture detailed information regarding psychological coping, physical performance with the prosthesis and patient perceptions across all areas of function. Furthermore, in aggregate, data from these measures has the potential to reveal trends in outcomes for different levels of amputation, different prosthetic options and provider care model effectiveness.

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PERSPECTIVES OF SUBJECTS WITH UPPER LIMB ABSENCE ON THE RISK FACTORS FOR MUSCULOSKELETAL COMPLAINTS.

Anneliek A. Peters, Sietke G. Postema, Michiel F. Reneman, Corry K. van der Sluis

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

ABSTRACT

Background: Musculoskeletal complaints (MSCs) are a highly prevalent problem in subjects with upper limb absence (ULA). Studies have been conducted to better understand the risk factors for the development and persistence of MSCs, and show relations with psychological and work-related factors. The opinions of patients with ULA have not been taken into account so far. Their perspectives can contribute to address important factors and aid in the improvement of treatments. This study therefore executed a focus group with subjects with ULA, to get insight in the patient perspectives and to develop a framework of all factors involved in the development and persistence of MSCs.

Methods: A focus group was held with adult individuals with ULA. With open questions, the general topic of MSCs and the main topic of the risk factors for MSCs were addressed. The transcript of the focus group was used to build a framework, by formulation (sub)categories of risk factors in an inductive way. The final set of categories was entered in the Atlas.ti software to identify sections of the transcript corresponding to a (sub)category.

Results: Eleven subjects with ULA participated in the focus group, of which three experienced no MSCs and eight had MSCs in the previous year. The opinions of the participants resulted in five main categories containing 29 subcategories: prosthesis-related, psychological & cognition, environment, general, and activities. Especially the factors in the ‘psychological & cognition’ and ‘activities’ category were deemed important.

Conclusion: The outcomes of the focus group regarding the categories ‘psychological & cognition’ and ‘activities’ cannot be endorsed with current literature, as literature on these categories is limited. Future research should therefore address the gaps between the patient perspectives and the literature, to fill the gap and to extend the knowledge on risk factors for MSCs.

INTRODUCTION

Many subjects with upper limb absence (ULA) complain about musculoskeletal complaints (MSCs), as we noticed in our clinic. Frequently heard complaints are carpal tunnel syndrome, epicondylitis (tennis elbow), stenosing tenosynovitis (trigger fingers), shoulder impingement, and neck and back complaints. Several studies have investigated the prevalence of MSCs in persons with ULA and the related population characteristics [1-4]. The year prevalence of MSCs in Dutch individuals with ULA was shown to be twice as high compared to their two-handed peers (65% versus 35%) [1]. MSCs are often chronic and observed in the residual limb, unaffected limb, neck or back, and with higher pain intensity and resulting in higher disability [1, 2]. Presence of MSCs in single-handed individuals will result in dual disability; adding disability due to MSCs to the disability caused by single-handedness [5]. This has consequences for daily life and emphasizes the high personal and societal impact [2, 3].

Studies have been conducted to better understand the risk factors for the development and persistence of MSCs. In patients with ULA, the presence of MSCs was associated with higher perceived physical work demands and lower general and mental health [1, 3]. Additional risk factors for the presence of MSCs in individuals with ULA are: higher age, being divorced or widowed, and lower mental health [1]. Prosthesis wear (daily duration of prosthesis use, number of activities with prosthesis use, and type of prosthesis) did not appear to be related to the presence of MSCs [1, 2, 4]. A possible explanation provided for the differences in prevalence between the subjects with ULA and the two-handed population, may be the use of compensatory movements, but this has not been examined so far [4].

The opinions of patients with ULA themselves on contributing and persisting factors for MSCs has remained underexposed at the moment. Their experiences with MSCs and their perspectives can highlight the importance of several factors, and can address the need for improvement of treatments. This study therefore executed a focus group with subjects with ULA, to get insight in their opinions and to develop a framework of all factors involved in the development and persistence of MSCs in this population.

METHODS

The Medical Ethics Review Board of the University Medical Center Groningen (METc UMCG) concluded that formal approval of the study was not necessary (METc 2019/228). All participants signed an informed consent before the start of the study.

Participants were recruited via a list of adult eligible patients composed by a clinician and via an advertisement in a magazine of the Dutch patient organization for persons with ULA. At the start of the focus group, which took place in April 2019 at the UMCG, the Netherlands, participants filled in a short questionnaire with socio-demographic data. During the 60 minutes-focus group, open questions about MSCs were asked. The first two questions introduced the topic of MSCs: 1) Who is familiar with MSCs, and if so what type of complaints have been experienced?; 2) Who is not familiar with MSCs, and how can that be explained? However, the main topic of this focus group were the risk factors of MSCs: 3) What are/could be the causes of these complaints?

The audio-recordings were transcribed verbatim. A framework was composed based on the transcript of the focus group. Two assessors (AAP, SGP) started with the familiarization process by reading the transcript of the focus group to get a sense of the whole text. Independently of each other, they developed a framework by formulating main categories and subcategories of risk factors in an inductive way. These (sub)categories were discussed and the data was reassessed. In a second discussion, differences were deliberated to reach consensus until a final set of categories was determined. In the next step, the transcript was entered in the Atlas.ti software. The framework was applied to the transcript and sections of the transcript that corresponded to a particular (sub)category were identified. The selected information was displayed in a list of quotes and corresponding categories. The results were analysed and discussed to draw conclusions until consensus was reached between the assessors.

RESULTS

Eleven participants (six males) participated in the focus group. Median age of the participants was 46.3 years (range 31.4 – 69.7 years). Three participants did not experience MSCs in the previous year, while eight did. Of these eight participants, seven experienced MSCs during the last four weeks. The median duration of MSCs was 3.5 years (range 0.5 – 20 years). Three participants perceived their pain as light, one had quite some pain, and one rated their pain as both these options (two subjects failed to fill in this question).

The opinions of the participants resulted in five main categories containing 29 subcategories (Fig. 1): prosthesis-related, psychological & cognition, environment, general, and activities. The two most mentioned categories were ‘psychological & cognition’ and ‘activities’. The main focus within the ‘psychological & cognition’ category was on the problem with setting boundaries. Participants felt that they had to give at least 150% in all of their activities, resulting in complaints: *“…, with what I do have, wanted to overcompensate. More in a way to prove: I can do everything. What you can do, I can do too. And then some extra.”* Difficulty to accept that they had a loss of function and, especially in those with acquired amputations, the wish to return to work in the same manner as before the amputation, contributed greatly to the development of MSCs. In the category ‘activities’, they addressed that compensation with the non-affected limb and performing physically demanding tasks were important contributors to the presence of complaints: *“I am missing my left hand and I, in order to compensate, perform everything with the right side. And that makes me really chronically overload my right shoulder. But also my head, neck, just the whole area. Sometimes I have, what do you call that stupid stupid thing, … a tennis elbow.”*

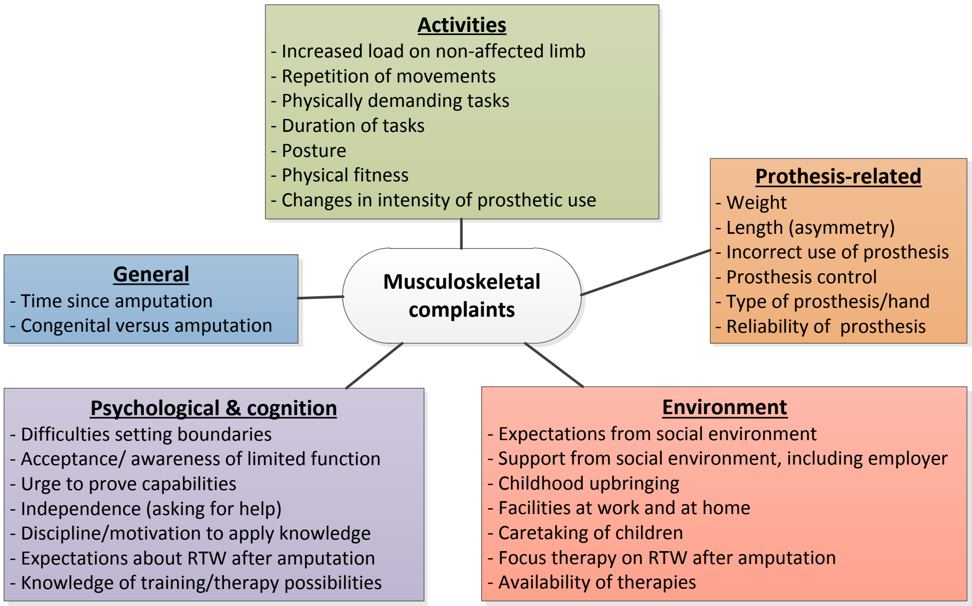

Figure 1: Framework of risk factors for the development and persistence of musculoskeletal complaints according to persons with upper limb absence. RTW: return to work.

DISCUSSION

A framework was made based on the opinions of patients with ULA on origin and maintenance of MSCs. This framework highlights themes that are risk factors for MSCs according to the patient population. Comparing this framework to several models [6-9], shows similarities with the International Classification of Functioning, Disability and Health (ICF) [9]. This framework will help to assess necessary treatment modalities of individuals with ULA, who experience MSCs. Using a distribution of risk factors comparable to the ICF-model is helpful to understand and measure consequences of MSCs, and can be used in clinical situations.

The opinions of the persons with ULA highlight the importance of psychological and physical factors. Psychological factors such as coping, support and work-related factors have been addressed in previous studies [1, 3]. On the contrary, studies about the physical factors have not been executed in subjects with ULA so far, even though studies mention compensation as a possible risk factor [4]. Investigating compensation strategies should be one of the research priorities for future studies.

Furthermore, to strengthen the framework, it should be supplemented with results of a literature review focusing on risk factors of MSCs in this population. Thus, creating an overview of all factors that may contribute to the development and persistence of MSCs. This overview may help synthesizing research priorities, which can then be taken into account in the development of new and better interventions to prevent and to treat MSCs in single-handed individuals.

In conclusion, patients suggest that psychological and physical factors play a major role in the development and persistence of MSCs. However, limited literature results are available to support these findings. Future research should examine the current scientific knowledge on MSCs in this population in order to complete the framework. Thereafter, discrepancies between patient perspectives and the literature should be addressed. Ultimately, more knowledge on population-specific risk factors of MSCs will allow to treat MSCs more effectively and reduce disability.

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PROSTHETIC REHABILITATION OF BILATERAL ANTECUBITAL PTERYGIA WITH CONCOMITANT CONGENITAL HAND DEFICIENCIES: A CASE STUDY

Anthony Berland CPO[1], Phillip M. Stevens Med, CPO[1,2]

[1]*Upper Limb Prosthetics Program, Hanger Clinic, San Antonio TX, USA*

[2]*Department of Clinical and Scientific Affairs, Hanger Clinic, Austin TX, USA*

ABSTRACT

This case study presents upon the unique application of prosthetic rehabilitation principles for a young man presenting with bilateral antecubital pterygia or webbing of the elbows. His case was further complicated by substantial bilateral congenital hand deficiencies. Prior to prosthetic intervention, this patient's upper limb function was confined to midline manipulation performed with his elbows. Following a surgical release of the webbing of his right elbow he had sufficient mobility to justify an exploration of prosthetic rehabilitation. We report on the initial prosthetic fitting which has substantially expanded this young man's working envelop and upper limb function. This was accomplished through dual-site direct myoelectric control of an electric hand mounted in relative internal rotation to facilitate midline function and preserve the patient's sensory input from his right residual forearm and hand.

INTRODUCTION

Antecubital pterygium syndrome has been defined as an extremely rare genetic disorder characterized by bilateral, fairly symmetric antecubital webbing extending from the distal third of the humerus to the proximal third of the forearm with associated musculoskeletal abnormalities. In the case in question, this physical presentation was further complicated by the presence of concomitant congenital hand deficiencies. Collectively, this left the patient largely bereft of meaningful upper limb function. This case describes the initial prosthetic fitting process and the subsequent improvements in upper limb functionality. Written informed consent was obtained for the presentation of this case study.

PATIENT PRESENTATION

The patient initially presented with bilateral webbing of the elbows restricting him to a non-functional 5-10° of elbow function (Figure 1). Distal to the elbow webbing, the patient presented with shortened forearms with significant congenital hand deficiencies inclusive of a single residual digit on his left upper extremity and two residual digits on his right extremity. Active wrist flexion and extension was present in both upper limbs (Figure 1,2). The patient's presentation was otherwise unremarkable with normal lower limb and cognitive function.

Figure 1: Left elbow and wrist mobility demonstrated in pictures of end range wrist and elbow flexion and end range wrist and elbow extension.

Following surgical release to the right elbow, the patient became capable of an expanded range of elbow motion, creating a possibility of additional functional gains through prosthetic rehabilitation (Figure 2).

Figure 2: Right elbow and wrist mobility demonstrated in pictures of end range of wrist and elbow flexion and end range of wrist and elbow extension

Prior to prosthetic intervention, the patient's upper limb function was largely confined to bilateral manipulations performed by the elbows, or using his residual hands to stabilize objects against his face to facilitate fine motor control. Writing, for example, was performed through neck movement with the residual limb stabilizing a pen against the head. Eating required the stabilization of a fork handle between the residual limb and head to stab a portion of food, followed by laying the fork on the table top where it could be bitten off of the fork.

PROSTHETIC MANAGEMENT

Recognizing the value of preserving the sensory input provided by the residual right forearm and hand, and the functional value of a mobile, sensate limb segment, the decision was made not to enclose the distal aspect of the limb within the socket. Further, recognizing the value of maintaining an anatomic length the prosthesis, the decision was made to mount the forearm, wrist and hand of the prosthesis in relative internal rotation. This was assessed dynamically in a test socket fitting prior to the completion of a definitive device (Figure 3).

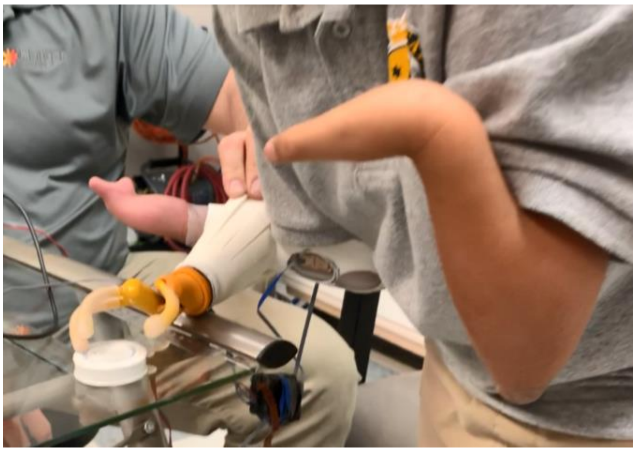


Figure 3: Dynamic assessment of the internally rotated prosthetic forearm to ensure optimal upper limb function.

Dual-site direct control was used to control prehension of a pediatric hand. EMG signals were obtained from the wrist flexors and extensors mounted within a costum silicone socket. The prosthesis also included an internal battery with an adjustable friction rotation wrist (Figure 4).

We subsequently observed that he needed increased wrist motion for different activities and installed a ball-in-socket universal friction wrist (myolino wrist 2000) to further increase his functional envelope. This increased his capacity to feed himself as well as write in a more ergonomic position.

Figure 4: Definitvie dual-site myoelectric prosthesis

The functional benefits associated with the prosthesis were immediately apparent as the child demonstrated the ability to grasp and lift objects from a table. He is now able to write with the aide of the prosthesis, sitting erect with no need to lower his face to the table top. Similar benefits have been observed with eating as this young man is able to grasp a fork and raise it to his mouth with no need to lower his face to the table top. The prosthesis has broadly enabled other midline functions (Figure 5).

Figure 5: Definitive prosthesis fully donned, demonstrating the capacity for midline function.

Future prosthetic plans include replacing the current terminal device to a more durable small adult hand. The surgical team is not currently considering a soft-tissue release on the left extremity due to concerns of possible secondary damage to the neurovascular bundle. We have considered the construction of a pass-through elbow-disarticulation style prosthesis on the left to facilitate bimanual function such as personal hygiene, riding a bike or sports activities, but this has not yet been formally explored.

CONCLUSION

This case illustrates a novel application of prosthetic rehabilitation principles. The case demonstrates a careful balance between providing a prosthetic enhancement to the affected extremity without overly compromising its native movement and sensory input. While the resulting device is less physiologic in its appearance, it meets the predominant needs of the patient in allowing him to write and eat in a more acceptable body alignment and position.

TAKE-HOME TRIAL OF THE GLIDE HAND AND WRIST MYOELECTRIC CONTROL ALGORITHM: A CASE STUDY

Chris Baschuk[1], MPO, CPO, FAAOP, Rahul Kaliki[2], PhD, Richard F. *ff.* Weir, Ph.D.[3,4],

and Jacob L. Segil[3,5], PhD

[1]Handspring Clinical Services, [2]Infinite Biomedical Technologies, LLC, the [3]University of Colorado Denver, [4]Rocky Mountain VA Medical Center, and [5]University of Colorado Boulder

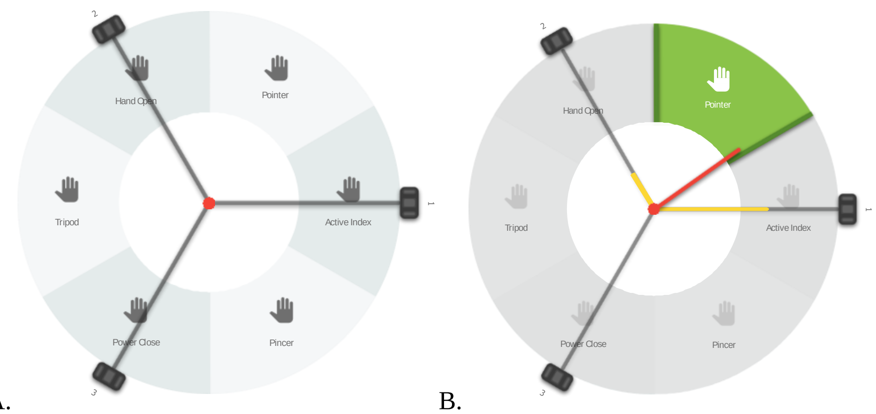
ABSTRACT

One of the most exciting developments in the field relates to the mechatronic advances which have enabled the creation of dexterous terminal devices, wrist rotators and powered elbows. However, their clinical impact has been limited by a lack of effective myoelectric control strategies. To address this challenge, we have developed a novel control strategy based on the Postural Control algorithm, which we call the Glide Controller. In this paper, we describe the first clinical fitting of the Glide system and present qualitative results on the fitting outcomes. We also discuss the implications of this control strategy from a patient and clinician perspective.

INTRODUCTION

The disparate progression of myoelectric control algorithms and their associated multi-functional prosthetic hands has caused mismatched technologies to become available to people with upper limb amputation. Several multi-functional myoelectric prosthetic hands are available today; however users are only able to access a small subset of the total number of grip patterns which are possible [1]. The prosthetic hands typically come with several methods for accessing different grip patterns including 1) myoelectric triggers, 2) buttons on the hand, or 3) gesture control [2], [3], [4]. These switching mechanism can require up to three different steps to switch between the current grip and the desired grip. Not surprisingly, many amputees find this process cumbersome and non-intuitive [5], [6], [7]. Moreover, using muscle triggers such as a co-contraction to control multiple grip patterns or movements is considered slow, cognitively demanding and unintuitive [5], [6], [7].

An intuitive control method, called pattern recognition, is emerging, however, several hurdles remain. Many researchers (including ourselves) have turned to pattern recognition of multichannel myoelectric signals in order to develop more intuitive control of advanced prostheses including the control of grasp patterns of multi-functional hands [8]. Pattern recognition algorithms seek to correlate patterns of surface EMG activity with a given intended movement command [9], [10]. Correlation is determined by calibrating a machine learning algorithm with labelled training examples in the form muscle activity recorded while the user holds a static posture. Because these patterns are representative of natural behaviors prior to amputation, control of the prosthesis via pattern recognition is intuitive and potentially increases the number of controllable DOFs. The most significant challenge for pattern recognition algorithms is that they require highly consistent and noise-free EMG signals [6]. This is



A. B.

Figure 1: A. Exemplary Glide domain where the EMG electrodes (1-3) are mapped in a radially fashion. Various hand grips are placed into wedges around the domain with a null state surrounding the origin (white). B. The real-time EMG activity is present with the yellow vectors and the resultant vector (red) determines the hand or wrist function that is selected.

particularly true as the number of degrees of freedom (DOF) in prosthetic hands increase. Thus, it would be highly preferable to develop a solution that can work without any calibration or extensive subject training.

Here we present an alternative control strategy, the *Glide* myoelectric control algorithm, which maps the electromyographic (EMG) signals to a radial mapping of prosthetic hand grips or wrist functions (Figure 1). The Glide algorithm is based upon our previous work on the Postural Control algorithm in the Biomechatronics Development Laboratory [11]–[13]. The basis of Glide algorithm is the vector summation of EMG signals from 2-8 EMG electrodes which is manifested as a “Glide vector” which is projected onto the Glide domain. The domain can be partitioned into “wedges,” which are correlated to single hand grips or wrist functions. A given wedge’s inner radius determines the onset threshold for a movement. Once the Glide vector exceeds the onset threshold, the amplitude of the Glide vector is proportionally mapped to the velocity of the wedge’s associated movement until the vector reaches the outer radius of the wedge, which corresponds to the maximum velocity of the movement. The mapping of the Glide domain is adjustable so that wedges can be placed anywhere in the domain, the inner and outer radius of the wedge can be independently changed, and the arc-length of each wedge can be made larger or smaller. These customizations allow for strong independent EMG signals to command certain functions and co-activity of other EMG signals to control other functions. The customizability ensures that the system can be fit to myoelectric prosthetic users with a broad range of abilities. Here we present a case study of a subject with trans-radial amputation who utilized the *Glide* algorithm with both hand and wrist function in a take-home trial.

METHODS

A single subject was recruited and informed consent obtained by clinicians at Handspring Clinical Service office in Salt Lake City, UT. The subject presented as a recent trans-radial amputee with a long residual limb length (8.5”). This subject was a novice myoelectric prosthetic user and had no prior experience with a myoelectric device outside of clinical sessions. The subject was originally amputated at a wrist disarticulation level but underwent a surgical revision for shortening to remove a neuroma and to improve the shape of the residual limb for prosthetic fitting. In addition, the surgeon salvaged and relocated the flexor pollicis longus muscle closer to the surface in order to provide an additional myosite for surface EMG control. The subject was fitted with a three-site *Glide* system where the electrodes were placed over the following muscles: 1) flexors digitorum, 2) extensors digitorum, and 3) flexor pollicis longus. The *Element* electrodes (Infinite Biomedical Technologies LLC, Baltimore MA) were integrated into the custom self-suspending HTV silicone prosthetic socket, the *FlexCell* battery and the *Glide* control system was integrated into the outer prosthetic socket. The TASKA prosthetic hand (TASKA Prosthetics, Christchurch, New Zealand) and wrist rotator (Motion Control, Salt Lake City, Utah) were utilized to provide the user with multiple hand grasps as well as wrist pronation and supination.

The *Glide* algorithm was configured to include the following hand grips and wrist motions: 1) hand open, 2) hand close, 3) wrist pronation, and 4) wrist supination. The gains for each of the electrodes were adjusted independently. EMG smoothing was enabled. A feature called walls was also enabled which prevents activation of a different hand/wrist function until the signal drops below the on threshold of the active Glide domain wedge.

After the subject enrolled in the study, the prosthetic system was fitted to the subject and tuned for best performance by the prosthetist. Training on use of the system was conducted by the prosthetist. The subject completed a battery of outcomes measures including 1) The McGann Feedback Form, 2) The OPUS: Satisfaction With Device and Services, 3) OPUS Upper Extremity Functional Status, and 4) OPUS: Health Quality of Life Index. The subject went home with the *Glide* system for a total of 4 weeks. Outcome measures were collected at initial fitting, two weeks post-delivery, and four weeks post-delivery. The outcome measure results and qualitative comments from the subject and prosthetist are provided here.

RESULTS

<u>Experimental Results</u>: The outcome measures were collected during the initial fitting, two-week session, and four-week session. Table 1 depicts the outcome measures over those sessions. The McGann Client Feedback Form results indicate an increase in prosthetic satisfaction across the four-week trial from 53% during the initial fitting to 97% satisfaction during the four-week session. The OPUS results provided a mixed description of the patient’s satisfaction, functional status, and health quality in that not all outcome measures improved across the four-week session. Nonetheless, the single-subject quantitative results for the first-time use of a new technology is an encouraging step

forward and suggests that the Glide algorithm can be an affective tool for the control of multi-functional prosthetic hands.

Table 1. – Outcome measure results across the initial fitting, 2-week session, and 4-week session

Experimental Session	McGann Client Feedback Form	OPUS–Satisfaction with Device	OPUS-Functional Status	OPUS–Health Quality of Life
Initial fitting	53%	39	46	74
2-week session	88%	36	30	78
4-week session	97%	38	43	73

Qualitative Results: The subject owns an excavation company and has a history of operating heavy machinery. This type of equipment utilizes joysticks with multiple switching mechanisms to manipulate the implements of the equipment. This experience was very useful for translating into prosthetic control. He was quoted as saying, "In the beginning I thought it was pretty easy. And the more and more as I go with this I recognize it is capable combining functions to do something different. So I'm getting better at it." His responses to the McGann Feedback forms indicated that as he became more familiar with the system his satisfaction increased. During the take home trial period the subject was fit with a Glide system with three electrodes. During one of the follow-up appointments he commented that he wanted to try adding a fourth electrode into the system as he stated, "I have the signals" referring to his ulnar deviators. This fourth electrode will be added in the future and further data will be collected.

DISCUSSION

Technological Progress: The Glide system is the next logical iteration of a traditional two site myoelectric control system. It has clinical implications for individuals who have had a conventional amputation surgery, but also has significant added benefits when combined with more contemporary amputation surgical methods such as TMR. From a clinical perspective it bridges the gap between a two-site myoelectric system and a full pattern recognition system. Selecting among multiple movements can be simpler than using EMG triggers such as co-contraction, double and triple impulses. While "joystick" control of the wrist is the most straightforward method to access different wedges within the Glide domain, it is also possible to use intuitive motions for control. It also bridges this gap from a cost standpoint as well. Fabrication is no more difficult or complex than a traditional two site system. The space requirements for the system are also minimal within the socket. Processing power consumption is low with no appreciable reductions in battery life as compared to a two-site system.

Clinical Perspective: There were some initial challenges in the clinical fitting as this was the first clinical application of the Glide algorithm. Part of the challenge was in learning how to refine and fine tune the arc lengths of the wedges, adjusting the gains and thresholds, enabling and disabling the walling features, and then proper queueing and instruction for the user. However, the technology proved to be quite adaptable and flexible. Initially the subject was sent home with only hand functions on the primary axis and the wrist functions as secondary fast rise actions like what a four-channel control system would be. This was challenging for him. Given his lack of myoelectric control experience he would inadvertently activate the wrist functions quite often, which proved to be frustrating to the participant. In particular when the wrist would start rotating unexpectedly and get into a unnatural anatomical posture, his whole ability to control the prosthesis would degrade. He expressed that this was because once the prosthesis was in an unnatural posture his sense of embodiment of the prosthesis completely disconnected. Fortunately, with an update to the control algorithm, wrist functionality was added as a primary function instead of as only a fast rise secondary function. Doing so allowed for defining additional Glide domain wedges for wrist control which the subject was able to activate with a high level of accuracy.

Initially when training the subject, he was queued to try and visualize moving his phantom limb as would be done in pattern recognition training and calibration. This worked to some degree, however it was never really consistent. The resultant vector would end up moving around quite a lot and not stay in one clearly defined area. It took some time to recognize that this queueing would not work and that another strategy needed to be developed. It was determined that we need to help the users conceptualize that the electrodes function somewhat like a joystick. It is best to put the primary functions right on the axis of the electrodes on the Glide domain. Once the subject has good control of each independent axis, then they can be queued to start trying to make combinations of contractions between adjacent electrodes on the Glide domain. The goal being to help the subject generate a resultant signal that is exactly in the middle of the pair of electrodes on the Glide domain. In doing so, with three electrodes six separate functions can be controlled. With four electrodes, eight separate functions could be controlled. By also allowing for fast and slow rise the potential exists to control double the amount of functions. The Glide system is not limited to the number of regions that could be created. Therefore, as an individual gains improved control in combining the signals, additional wedges can be added to the Glide domain to add additional functions. This will allow the individual to be able to access specific grip patterns of multi-articulated terminal devices as well as additional wrist functions such as flexion and extension.

Unlike in pattern recognition systems where the process of classification is somewhat obscured from the user and the prosthetists, the Glide system allows the prosthetists and the user of the technology to visually see on the Glide domain what the function will be without any ambiguity or uncertainty. It also allows the prosthetist to easily adjust the Glide domain mapping and ensure easier selection of each function. By increasing the arc of the wedge on the Glide domain and adjusting the on and maximum thresholds, the prosthetist can effectively accommodate for accuracy and fatigue of signals. Clinically, it was found to be very helpful to be able to adjust this tolerance. Past clinical experience with pattern recognition systems has shown that sometimes throughout the day as a user's muscles fatigue their classification accuracy may diminish resulting in unwanted behavior. The Glide domain interface allowed for the clinicians to adjust the wedge size and shape in order to avoid this pitfall.

Clinical Implications: A system built on the Glide algorithm provides a novel advance to traditional myoelectric control. When set up with only two electrodes it functions in the same way that a conventional two site system would. However, it provides a significant clinical advantage for controlling an increased number of functions and motions of a prosthesis when additional electrodes are added into the system. This is accomplished without time consuming additional fabrication and minimal additional hardware and processing power. Increasingly, the possible controllable motions of a prosthesis outnumber the inputs that a user has available thereby requiring complex switching strategies or signal processing algorithms in order to activate them. The limitation on a user's ability to benefit from these additional motions is correlated to the number of inputs available to them. Future applications could see connecting non-EMG inputs into the Glide system in combination with EMG signals. This could help individuals with limited surface EMG sites, such as higher level amputees, also benefit from this technology.

Because the Glide system allows for more granular control, the amount of time programming was longer than for a conventional two site system or for a pattern recognition system. There is a learning curve to the system, but over time with further fittings and documentation of outcomes a guideline of best practices will be able to be developed. This will be critical for widespread adoption by clinicians.

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USING PATTERN RECOGNITION TO ENHANCE PROSTHETIC CONTROL IN PATIENTS WITH PROXIMAL AMPUTATIONS WITHOUT TARGETED MUSCLE REINNERVATION: A CASE SERIES

Anthony Berland CPO[1], Phillip M. Stevens Med, CPO[1,2]

[1]*Upper Limb Prosthetics Program, Hanger Clinic, San Antonio TX, USA*

[2]*Department of Clinical and Scientific Affairs, Hanger Clinic, Austin TX, USA*

ABSTRACT

The relatively recent commercialization of pattern recognition has occurred simultaneously with the proliferation of Targeted Muscle Reinnervation (TMR). Reports on applications of pattern recognition have generally been its application on proximal amputation post-TMR procedures or on transradial amputations in the absence of TMR. This case series highlights two successful applications of pattern recognition to patients with high level amputations who had not undergone TMR. In both cases, the users experience enhanced prosthetic control with reduce frustration and cognitive burden of prosthesis use. Pattern recognition appears to be a viable control strategy in high level upper limb amputation without TMR procedures.

INTRODUCTION

Direct control systems have been the traditional standard for myoelectric control of upper limb prostheses. In dual-site direct control a pair of surface electrodes are positioned over a set of antagonistic muscles with distinct EMG signals from these muscles providing threshold-based, proportional control of opposing prosthetic movements. However, the muscular actions of the controlling EMG sites are often physiologically inappropriate and counterintuitive with respect to the desired prosthetic movements [1]. This is more pronounced at high-level amputations where the muscles of the upper arm and shoulder girdle are recruited to control hand prehension and wrist rotation. Further, with direct control systems for high-level amputations the number Of EMG control inputs for prosthetic movements are insufficient, often requiring the user to generate specialized EMG signals to cycle between joint segments of the prosthesis [1].

In contrast to direct control, pattern recognition control reads EMG information from throughout the residual limb. Prosthetic control is provided through the recognition or correct classification of collective muscle patterns obtained from throughout the limb. This allows for direct control of multiple prosthetic movement patterns.

The commercialization of pattern recognition has occurred simultaneously with the proliferation of TMR, an innovative surgical procedure designed to increase the number of independent EMG sites available upon a residual limb. Publications on the application of pattern recognition in prostheses for high-level amputation have generally been confined to individuals who had undergone TMR procedures [2-5]. This cases series will highlight two successful applications of pattern recognition for high-level amputations that have not been revised using TMR techniques. Written informed consent was obtained from both case subjects.

SHOULDER DISARTICULATION CASE

DV presented with a right shoulder disarticulation amputation (Figure 1). He was initially fit with a passive prosthesis to restore an aesthetically acceptable appearance in community activities.

Figure 1: Right Shoulder disarticulation

A year later he was provided with a second prosthesis. The EMG signals on DV's chest wall were so strong that they effectively drowned out the more modest EMG signals that could be obtained from his upper back. As a result, the control strategy of this first electric prosthesis was a single-site direct control.

More specifically, the EMG signals derived from his chest wall were used to control the sequential movement of his elbow, wrist and hand. EMG signals exceeding the 1st level threshold provided control input to the joint under active control using an alternating strategy in which a brief latency period between contractions allowed control to switch to movement in the opposite direction (I.e., from an elbow flexion to an elbow extension command). Brief spikes exceeding a 2nd level threshold acted as the sequential switching signal between joints.

DV wore this system regularly and became quite adept at its use, but frequently commented on the tedious nature of its control which could become frustrating in the execution of finer motor movements.

Several years later, soft tissue revisions to the limb required the replacement of this prosthesis. At that time pattern recognition was assessed as a possible means of enhancing prosthetic control. During this assessment it was discovered that while the signals were dwarfed by the more powerful signals of the chest wall, discernible EMG signals could be obtained from the infraspinatus, supraspinatus and latissimus dorsi. While these signals were inadequate to exceed the threshold requirements of direct control, and could not be adequately separated from EMG activity of the pectoralis major, they were sufficient to inform the nuanced patterns required in pattern recognition.

A dynamic test socket was constructed with a single pair of anterior electrodes and 3 pairs of posterior electrodes located over the targeted muscle bellies (Figure 2). Over several weeks of use, the DV was able to consistently generate distinct signals for elbow flexion and extension, wrist pronation and supination and hand opening and closing.

Figure 2: Dynamic test socket with 8 electrodes positioned over targeted muscle sites

This control strategy was preserved in the fabrication of the definitive prosthesis, inclusive of an Espire Elbow, Motion Control wrist rotator and BeBionc hand (Figure 3). Passive grip selection using the contralateral hand provided the patient access to 8 distinct grip patterns.

Figure 3: Definitive prosthesis inclusive of pattern recognition control of an Espire Elbow, MC wrist rotator and BeBionic Hand.

While DV continues to prefer his passive prosthesis for much of his community activities, he wears the more advanced arm regularly to accomplish basic ADLs around the house with a specific interest in meal preparation. He is extremely pleased with the enhanced control and reduced frustration in operating the system.

SHORT TRANSHUMERAL CASE

KA presented as a legacy user of a range of upper limb prostheses following his short transhumeral amputation secondary to an IED blast sustained in combat (Figure 4).

Figure 4: Short transhumeral amputation secondary to IED blast injuries

At the time of KA's presentation to our clinic, he was using a hybrid prosthesis with a body-powered elbow, passive control of pronation and supination and dual-site direct control of a BeBionic hand. He presented with ample

EMG signal from his residual triceps, but extremely weak EMG from his residual bicep. He was able to cycle between 3 targeted grips using an "open-open" switching signal, but his ability to consistently close the hand was poor. He expressed frustration with both his consistency of operation and the cognitive burden of prosthetic control.

In response to these deficits, pattern recognition was explored as an alternate means of myoelectric control. Eight electrodes were positioned over the anterior, medial and posterior aspects of the socket (Figure 5). These produced an extensive EMG palate that ultimately generated discrete control of active pronation and supination, hand opening and 3 discrete closing signals for his TASKA hand including general grasp, flexi-tool and a custom grip that allows him to hold his tablet while working as an environmentalist in a mine.

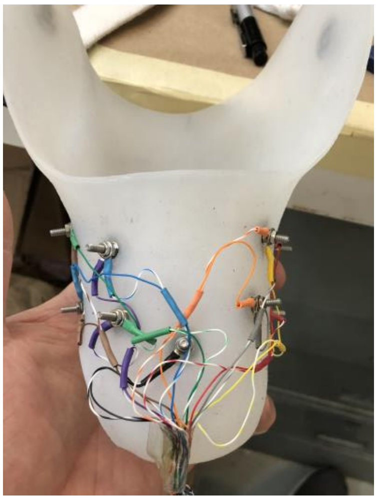

Figure 5: Placement of 4 pairs of electrodes to inform the patient's pattern recognition control scheme.

The patient's definitive hybrid prosthesis was inclusive of a suction socket with a body powered hybrid elbow and myoelectric control of a powered wrist rotator and a heavy duty multiarticulate TASK hand (Figure 6). The patient reports daily use of this prosthesis with specific application in his work setting.

Figure 6: Definitive hybrid prosthesis

CONCLUSION

Pattern recognition in the control of prostheses for high level amputations has largely been described in patients who have undergone TMR to expand the strength and availability of EMG control signals. In this case series we describe two high level patients who experienced substantial improvements in their control of their electric prostheses with the introduction of pattern recognition without the benefit of TMR. The ability of pattern recognition to recognize subtle distinctions in EMG patterns at proximal amputation levels appears to be sensitive enough to provide many discrete signal inputs even in the absence of TMR.

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USING DIAGNOSTIC ULTRASOUND FOR FITTING MYOELECTRIC PROSTHESIS IN INFANTS WITH CONGENITAL UPPER LIMB DEFICIENCY

Sayaka Fujiwara[1)], Chika Nishizaka[1)], Hiroshi Mano[1) 2)], Satoshi Endo[1)],
Kengo Ohnishi[3)], Nobuhiko Haga[1)]

1)University of Tokyo Hospital,2) Shizuoka Children's Hospital, 3) Tokyo Denki University

ABSTRACT

When fitting an infant with the first myoelectric prosthesis, finding an appropriate location and orientation for the myoelecrode to gain adequate signal at muscle contraction is troublesome.

Palpating the small muscle bulge in the short residual limb at short contraction without interactive communication is not easy and searching the location on the arm moving the sensor while watching the myoelectric signal on the screen is an overloaded task.

Diagnostic ultrasound B-mode image is a convenient and safe technique to visualize normal and pathological muscle and other anatomical variety in real-time in a non-invasive manner.

Since 2017, we are successfully using ultrasound diagnostic device (UD) for arranging the electrodes. UD is effective and useful at practice situation, since the dynamic visual feedback of the muscle contraction allows to easily and reliably locate the electrode location over the target muscle.

Diagnostic ultrasound should also be a good visual feedback system to train or detect proper signal strength from limb deficiency patients.

BACKGROUND

Congenital upper limb deficiency (CULD) is a rare disease which impairs both function and appearance of the limbs. The treatment approaches vary according to the type of deficiency. To provide better evidence-based medical care, it is necessary to establish the standard treatment of CULD. (1)

We established Limb Differences/Amputee Clinic since 2013. This outpatient clinic is not only for adult patients but also for children with congenital and acquired amputations or limb deficiencies. Our team members include rehabilitation physicians, occupational therapists, physical therapists, prosthetists, engineers and other comprehensive care members such as orthopaedic surgeons and paediatricians

We provide rehabilitation therapy including prosthetic interventions, such as conventional prostheses, myoelectric prostheses and so on.

Since 2017, we had started using Ultrasound diagnostic device (UD). UD is a convenient and safe technique to visualize normal and pathological muscle and other anatomical variety as it is not invasive and real-time. When fitting an infant with the first myoelectric prosthesis, we need to find an appropriate place for the myoelecrode to get a proper signal when the muscle is contracted.

The prescription of prostheses and proper fitting of a prosthesis require an adequate length of the stump. However with infants with short stump of transradial deficiency, we always have trouble finding an appropriate location and orientation for the myoelecrode to gain adequate signal at muscle contraction is troublesome.

Palpating the small muscle bulge in the short residual limb at short contraction without interactive communication is not easy and searching the location on the arm moving the sensor while watching the myoelectric signal on the screen is an overloaded task. UD is so convenient and useful at practice situations because we can place an electrode on the exact muscle inside of the infant's short stump with assurance.

CASE PRESENTATION

The subjects are three infants before 2-year-old of age with transradial limb deficiencies in the short residual limb with the first myoelectric prosthesis. (Fig.1)

When we prescribe and fit an infant with the first myoelectric prosthesis, finding an appropriate and suitable place for the electrode to get a proper signal is required when the the small muscle bulge of stump is contracted.

We use the diagnostic ultrasound system SNiBLE by Konica Minolta or finding appropriate muscle in the stump. Diagnostic ultrasound B-mode image is a convenient and safe technique to visualize normal and pathological muscle and other anatomical variety such as joint instability or laxity and muscle deficiency in real-time in a non-invasive manner.

finding an appropriate location and orientation for the electrode to gain adequate signal at muscle contraction is troublesome. It is often happening and we face the difficulty that finding the appropriate location and orientation for the electrode without interactive communication. This is because of the small muscle of residual limb and the thick subcutaneous fat. However, UD is an effective and useful device at practice situation such as the infant with a short stump of transradial deficiency, because we can find the place an electrode on the exact muscle in the short residual limb of the infant with assurance. The information and data from UD is not only reveals the presence of the muscle but the direction of muscle fiber as well. (Fig.2)

When fitting a toddler or young child with a myoelectric prosthesis, it is said that finding appropriate muscle sites to place an electrode is generally easy.（2） However it is not so easy with an infant around age 0 to 1-year-old who has the short residual limb. Finding small muscle bulge and direction of muscle fiber is nearly impossible to find on the infant residual limb because it is too small and difficult to see, due to no functional motion of the joint and thickness of subcutaneous fat.

Using diagnostic ultrasound B-mode image is effective technique for us to place the electrode on the infant small stump. All the three infant with short residual limb are easily successed the site selection, and they control with sure and good to operate the myoelectric.

CONCLUSION

The infant with a short residual limb of transradial deficiency, we always have trouble to find the ideal place for the electrode. It is also said that UD appeared to be more sensitive in detecting EMG and clinical observations, because it can visualize a large muscle area and deeper located muscles. (3)

We had started using UD. It is so convenient and useful at practice situations because we can place an electrode on the exact place of the muscle inside the infant short residual limb. This is where we can place the electrode by diagnostic ultrasound without a doubt or hesitation if the infant cannot contract the muscle in the short residual limb during site selection to control myoelectric.

Furthermore, the child can see their muscle contraction or movement by UD, we are confident that diagnostic ultrasound also becomes a good visual feedback system to train or detect proper signal strength from affected limb for young children and older who can recognize the image.

This report was approved by the Ethics Committee of the Faculty of Medicine of the University of Tokyo ; ethical approval number: 2373. At the University of Tokyo Hospital, all patients were given an explanation regarding personal information protection, including the use of clinical data for research.

FIGURE

Figure 1: Right transradial CULD short stump of 1-year old girl.

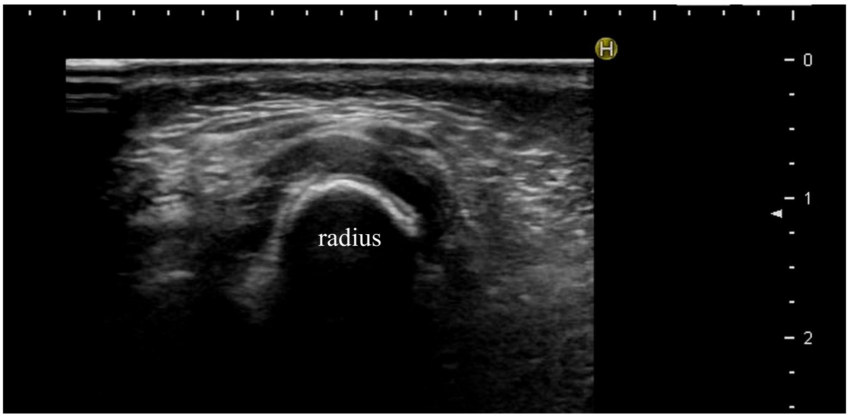

Figure 2: Cross section of CULD proximal stump by UD. Supinator muscle and other extensor muscles on the radial side of forearm. The UD can detect and visualise the muscle less than 2mm thickness.

ACKNOWLEDGEMENTS

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Video Games: Client Centered Adaptations

Intent: *To provide information and resources for clients with amputations and for providers, in order to return to video gaming.*

Video games have been a part of modern society since the late 1970s and 1980s. The people who played the early systems are now growing older, but many continue to play. Video games have also become a part of modern culture with the rise of the professions of eSports, online streaming videos, as well as in movies and television. Billions of dollars are spent within the video game industry to develop the latest and greatest systems and games. However, only recently have industry leaders created adaptive devices for gamers with disabilities.

Off the Shelf Adaptations

Microsoft recently released an adaptive controller that works with the Xbox One system and a Windows based computer. This is a first in the industry, a large company taking the initiative to market and sell an adaptive gaming product. The device has worked well, but for some gamers with an amputation, it has not been the best solution. The device starts at $99 USD, but may require the gamer to buy switches and other control devices. A standard Xbox One controller has ten buttons, and the adaptive controller only has two built in programable buttons. The gamer must acquire extra switches to account for those buttons, and at a cost of at least $45 USD per switch, the cost can easily build up. Another issue for the Microsoft device, is that it only works with Microsoft systems. This can be a challenge since many gamers are usually very loyal to their favorite brand, such as the Sony Playstation or Nintendo Switch, usually purchasing every generation of the brand's systems as they get released. Gamers may invest thousands of dollars buying games that only work for their system, but if they have an amputation, they may not be able to play their non-Microsoft system. There has been some success modifying the Microsoft controller onto other systems through adaptors and software, but this can have another increase in cost and can be more technical than some clients and/or providers wish to attempt.

Another type of controller that is commercially available are customizable controllers such as the *Scuf Controller* by Scuf Gaming International LLC. Typically, a popular choice for "hardcore" or professional gamers, these controllers have customizable buttons on the back of the controllers, built-up joysticks, and more software options. With these customizations, some clients with unilateral amputations or digit amputations have found that they are able to return to gaming. These controllers can work for the Sony Playstation, as well as the Microsoft Xbox. However, they can be expensive starting at $150 USD, and can require extra purchases for further customizations.

Non-profit organizations like Warfighter Engaged, Inc. out of New Jersey, have been building custom controllers and adaptations for gamers with amputations since 2012. Their focus is to get the gamer back into the game by adapting the environment with controller modifications and prosthesis enhancements. This has been successful at getting players back into the game.

In-Clinic Adaptations

Some clients are apprehensive about spending the amount of money required for some of the off the shelf adaptions. Therefore, with some in-clinic resources and a little ingenuity, low cost adaptions can be made.

Using thermoplastic material, joysticks on the standard controller can be built. Some patients have even been able to put long screws into the controller's joystick to increase their ability to reach it.

Typically, thumbs are required to manipulate the two joysticks on most controllers, which is difficult with a thumb amputation. By custom fabricating a thermoplastic prosthetic thumb, which are often used for early prosthesis training, clients can return to gaming. It is also recommended to add friction material, such as *Dycem*, to the end of the thermoplastic prosthetic thumb to provide a better grip to the device and controller.

Games for Therapy

There is no denying the therapeutic benefit of many video games. The Nintendo Wii continues to be used in various therapy settings across the world, even though the system was discontinued in 2013. However, many clients are more into the newest systems. There are new games for the current systems that also have therapeutic benefit. Nintendo has released a new Fitness game called *Ring Fit Adventure* for their Switch system. This requires the player to perform upper and lower body exercises to progress through the game and storyline. Games such as the *Just Dance* and *Dance Dance Revolution* require the player to use balance and whole-body movement to achieve in-game goals.

Another new frontier for therapeutic gaming is the increased availability of virtual reality (VR). VR systems have become more affordable, which enables them to be used in clinics. Stand-alone systems such as the *Oculus Quest* by Facebook Technologies, LLC is only $399.00 USD, but has everything included. No need to purchase another system. This allows clients to place themselves into an immersive virtual environment, and to safely attempt tasks they never thought possible.

Tips for Non-Gamer Providers

Some therapists and providers without gaming experience may be overwhelmed with new video game technology. However, gaming is just like other leisure pursuits, and can be analyzed just like any other activity. Gamers usually like to talk about their gaming systems and want people to play with them. Don't be afraid to ask the client questions about how to play the game or even try it out. Older systems such as the original *Nintendo Entertainment System* and Sega *Genesis* have been remade and are a great introduction to gaming, which can then be used to better relate to gamer clients. There are also many online resources with step-by-step instructions on most games.

So Game On!

Track: Clinical Research Studies

THE ABILITY TO PARTICIPATE IN SOCIAL ROLES AND ACTIVITIES IN WEARERS OF UNILATERAL TRANSRADIAL PROSTHESES

Phillip M. Stevens, MEd[1,2] Mandi M. Laurie, MS[1,3], Shane R. Wurdeman, PhD[1,4]

[1]*Department of Clinical and Scientific Affairs, Hanger Clinic, Austin, TX, USA*
[2]*School of Medicine, University of Utah, Salt Lake City, UT, USA*
[3] *Soules College of Business, University of Texas, Tyler, TX, USA*
[4]*Department of Biomechanics, University of Nebraska, Omaha, NE, USA*

ABSTRACT

The literature related to upper limb prosthetic rehabilitation has largely focused on body structure and function. The constructs of activity limitation and participation restriction are comparatively under represented. The intent of this effort was to assess the related constructs of activity and participation among a cohort of individuals using unilateral transradial prostheses and correlate these findings against measures of upper limb function, satisfaction, quality of life, prosthetic wear time and pain interference.

We observed the strongest correlation of patient reported activity and participation to be an individual's perceived bimanual upper limb function as measured by a custom PROMIS short form (r=0.74). Additionally, strong correlations were observed between activity and participation values and perceptions of both quality of life (r=0.44) and satisfaction with life (r=.37). The additional constructs of pain interference (r=.34) and reported prosthesis wear times (r=.32) also demonstrated weaker correlations with activity and participation.

INTRODUCTION

The World Health Organization's (WHO) International Classification of Functioning and Health (ICF) facilitates a comprehensive understanding of the challenges faced by individuals coping with illness or disability. As with many other physical disabilities, upper limb amputation is associated with immediate and profound impairments within the realm of body functions and structures. However, the ICF model encourages additional consideration of both activity limitations and restrictions to participation [1].

Gallagher et al identified frequently encountered activity limitations for this population. These included getting dressed (52.9%), taking care of household responsibilities (52.9%), and day-to-day work/school activities (40.0%) [2].

Additionally, in consideration of restrictions to participation, the most frequently identified restrictions have been suggested in employment or job seeking (91.7%), family life (41.2%), leisure/cultural activities (41.2%), sports or physical recreation (38.5%), shopping (35.3%), living with dignity (35.3%) and socializing (23.5%) [2].

In addition to the disability considerations identified within the ICF Model, Wurdeman, Stevens, and Campbell have reported upon the increases in individual quality of life and satisfaction associated with the use of lower limb prostheses [3]. Of particular interest to the present study is whether this relationship holds true for a population of upper limb prosthetic users as well, and what correlations may exist between perceived activity and participation and reported satisfaction and quality of life.

In 2004, the National Institutes of Health (NIH) launched the Patient Reported Outcomes Measurement Information System® (PROMIS®) "Roadmap Initiative" [4]. This effort sought to leverage modern psychometric techniques to improve the measurement of symptoms and health outcomes by generating and refining item banks across a range of health-related constructs. The initiative ultimately created numerous patient-reported outcome measures covering a wide range of both symptoms and functionalities as well as establishing a standardized scoring framework that could be used across illnesses, chronic health conditions, and the general population [4].

Among these instruments are a small series of short forms addressing an individual's perceived ability to perform their usual social roles and activities, appropriately entitled the PROMIS® Ability to Participate in Social Roles and Activities (PROMIS-APSRA). This construct aligns well with the considerations of activity limitation and participation restriction proposed by the ICF and has been assessed and published across a range of illnesses and disabilities.

Also within the available PROMIS measures is the PROMIS Physical Function Upper Extremity measure or PROMIS®-UE. This is a measure of an individual's perceived ability to complete tasks that require the use of one's upper limb. The PROMIS®-UE utilizes item-response theory to generate a probability-algorithm to measure both the response to an individual question, as well as the concurrent relationship between said item and domain specific items This provides a more advanced and holistic view of an individual's physical function both in terms of the response to an individual item and the relationship of the item to the entire measure.

In the case of the customizable PROMIS®-UE test bank, items can be selected to provide a psychometrically sound representation of an individual's overall perceived function level, rather than simply a measure of an individual's ability to perform a given isolated task. Item-response theory algorithms reduce the required number of items that must be administered while maintaining test validity, thus reducing the time burden for both the clinician and test taker.

The purpose of this paper is to report upon the ability of individuals using upper limb prostheses to participate in social roles and activities and those factors that may be closely correlated to this construct. Examined factors include upper extremity function, hours of wear time, quality of life, and post amputation satisfaction with life. A relationship between an individual's APSRA and upper extremity function and was hypothesized. Participation and activity was further hypothesized to be related to quality of life and post amputation satisfaction with life. Finally, hours of prosthesis wear time was hypothesized to be related to APSRA.

METHOD

Study design

During routine patient care, patient outcomes were collected from patients receiving maintenance or replacement of an upper limb prosthesis. The present data represents a multi-site review of all outcomes collected from May 2017 through December 2018.

Participants

Of particular interest to the present study were individuals age 18 and older, with unilateral transradial amputations, who were actively using any type of prosthesis.

Procedure

To assess upper limb function in the sampled cohort a custom 9-item short form from the PROMIS®-UE v2.0 item bank consisting of bimanual activities was administered (PROMIS-9 UE). Patients were asked to report the level of difficulty associated with each item using a Likert scale ranging from 1 (unable to do) to 5 (without any difficulty). Items included such tasks as opening and closing a zipper, cutting food using utensils and lifting or passing heavy items.

Bimanual activities were intentionally chosen to attempt to isolate those activities where prostheses would be more likely to influence upper limb function. Raw scores were converted to t-scores using the healthmeasures.net scoring service such that a score of 50 corresponds the average scores of the United States population.

Additional survey items included the 4-item short form of the PROMIS-APSRA. This construct aligns well with the considerations of activity limitation and participation restriction proposed by the ICF. Using the Prosthesis Evaluation Questionnaire-Well Being [5], patients rated their satisfaction with life (SAT) and quality of life (QOL) over the prior 4 weeks. Scored range from 1 to 10, with higher scores indicating higher levels of well-being.

Patients were additionally asked about their prosthesis wear times: number of days per week or month and number of hours per day.

Analysis

Data were initially reduced with only individuals that met inclusion/exclusion criteria included. The resulting data were then assessed using Pearson Product-Moment correlations and group means. Of interest were the relationships among the variables. Therefore separate Pearson correlation coefficients were calculated for PROMIS-9 UE t-scores, PROMIS-APSRA t-scores, QOL, SAT and reported daily prosthesis wear times. Correlations were calculated at the 95th percentile, after 1000 repeated bootstrap iterations. Standard Cohen [6] effect sizes were used to assess the Pearson correlation coefficient effect size.

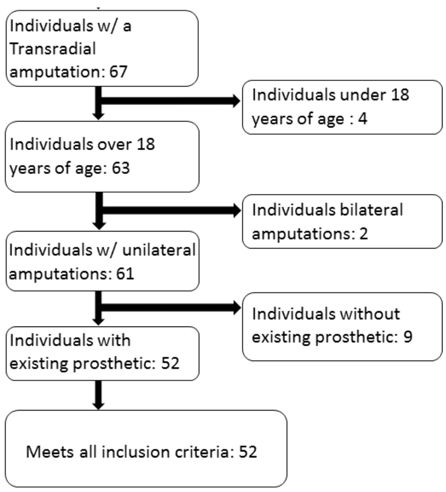

Figure 1: Data reduction flow chart resulting in 52 participants

RESULTS

Participants

Data were extracted from 67 patients that completed the outcomes assessments while visiting participating clinics. This number was further reduced to 52 users of TR prostheses (Figure 1). Demographic data is presented in Table 1.

Table 1: Patient demographics

Age (years)	48.9 ± 15.8
Gender	38 men; 14 women
Height (cm):	177.3 ± 9.9
Weight (kg):	85.7 ± 26.5
Time since amputation	44 ± 20.1 months
Reported prosthesis use	hours/day: 11.0 ± 5.0 days/week: 5.7 ± 2.3

Results

There was a large, significant, and positive correlation between the PROMIS-APSRA and the PROMIS-9 UE (r=0.738, Table 2). A large but lesser correlation was also observed between the PROMIS-APSRA and QOL (r=0.443) There were significant medium positive correlations between the PROMIS-APSRA and SAT (r=0.369), Pain interference (r=0.340), and reported daily prosthesis utilization rates (r=0.323).

Table 2: Correlation Coefficients

Variable	PROMIS-APSRA
PROMIS-9 UE	.738**
QOL	.443**
SAT	.369*
Pain Interference	.340*
Px Hours/Day	.323*

* Correlation is significant at the $p < 0.05$ level.
** Correlation is significant at the $p < 0.01$ level.

Discussion

The purpose of the present study was to assess correlates to reported participation in social roles and activities among individuals using a unilateral TR prosthesis. We observed percieved activity and participation values to correlate with reported bimanual upper limb capacity, reported quality of life and satisfaction with life, reduced reported interference from pain and increased prosthetic wear time.

We note that the strongest correlate to higher reported activity and participation scores was greater perceived bimanual capacity. Given the bimanual nature of the tasks in the PROMIS-9 UE and the inclusion criteria of an individual's active use of a prosthesis, we can reasonably assume that these scores represent the abilities of the sampled individuals to engage in bimanual activities with their respective prostheses.

Patients who exhibited the lowest level of perceived difficulty performing a range of bimanual tasks were found to have the highest PROMIS-APSRA scores. Further, PROMIS-9 UE scores correlated to PROMIS-APSRA scores much more strongly than to such variables as pain interference or reported prosthetic wear time. Accordingly, it would appear that the ability to perform a desired task when needed may be more closely related to activity and participation than either pain or prosthetic wear time.

The present paper suggests that the role of the clinical team is not limited to fitting an ideal prosthetic device but ultimately to ensuring that this device and the associated training facilitates the individual's ability to perform bimanual tasks when required.

The associations observed in the present study represent a starting point in connecting the goals of the clinical and rehabilitative teams, recognizing that upper extremity physical function is strongly connected to social function. Further, social function appears to be strongly connected to perceived quality of life, and moderately connected reported satisfaction with life. However, future studies are needed to further solidify these findings and better understand the influencing factors of these connections.

Limitations

The previous results notwithstanding, it is important to note a limitations of data collected at multiple sites. It introduces a potential for human error as data was collected by multiple clinicians. However, steps were taken to alleviate this error potential as clinicians completed training sessions as part of broader, ongoing clinical outcomes data collection training.

Further, we note that our observations were exclusive to users of unilateral transradial prostheses. The extent to which these observations may translate to individuals with more proximal amputation levels is not clear from the current data set. Additional, no analyses of prostheses type on the observed variables was performed.

Conclusion

Among the sampled cohort, among users of unilateral transradial prostheses, the strongest predictor of an individual's reported ability to participate in their social roles and activities was their perception of their upper limb physical function across a range of bimanual tasks. Not surprisingly, activity and participation scores were also strongly correlated with both the QOL and SAT scores reported by the participants. While the additional constructs of prosthetic wear time and pain interference demonstrated moderate correlations with activity and participation scores, these correlations were much weaker than those observed with an individual's perceived capacity to accomplish bimanual tasks.

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COGNITIVE LOAD IN LEARNING TO USE A MULTI-FUNCTION HAND

Helen Lindner[1], Wendy Hill[2], Liselotte Norling Hermansson [3, 4] Achim J. Lilienthal[5]
[1]*School of Health Sciences, Örebro University, Örebro, Sweden*
[2]*Institute of Biomedical Engineering, UNB, Fredericton, Canada*
[3]*University Health Care Research Centre, Faculty of Medicine and Health, Örebro University, Örebro, Sweden*
[4]*Dept. of Prosthetics and Orthotics, Faculty of Medicine and Health, Örebro University, Örebro, Sweden*
[5]*School of Science and Technology, Örebro University, Örebro, Sweden*

ABSTRACT

Despite the promising functions of a multi-function hand, it is challenging to learn to use a hand that has up to 36 grip patterns. If it requires too much cognitive load to learn to operate a prosthetic hand, the user may eventually stop using it. Measurement of cognitive load while learning to use a bionic hand will help the therapist to adjust the training pace and help the user to achieve success.

An innovative, non-obtrusive method for measuring cognitive load is by tracking eye gaze. Gaze measures provide pupil diameters that indicate subjective task difficulty and mental effort. Three subjects wore a pair of Tobii eye-tracking glasses during control training and performed eight activities. Eye-tracking data were imported in Tobii Pro Lab software for extracting pupil diameter during the activities. Pupil diameter (normal range: 2-4mm during normal light) was used to indicate the amount of cognitive load.

Pupil diameters were below 4mm in 9 out of 23 training activities. Pupil diameters were above 4mm in all three subjects when they used precision pinch to perform the activities “stack 4 1-inch wooden blocks” and “pick up small objects”. Subject 3 had pupil diameters over 4mm in all training activities. Pupil diameters were largest when the subjects were adjusting the grip and when they had difficulties in initiating the grip.

It seems appropriate to introduce no more than four grips during the first control training session. Further study is required to determine if pupil diameters will decrease over time when adequate prosthetic training is given.

BACKGROUND

In recent years, prosthetic technology has advanced significantly and many new hands with increased dexterity and functionality have been introduced to the commercial market. Clinicians want to offer the most useful device for their clients, however, it is challenging to learn how to operate a hand that has up to 36 grips. The cognitive load required to learn to use these hands and switch between the multiple grip patterns is unknown.

During training, most occupational therapists introduce features of these hands gradually so as not to overwhelm the client. As the client masters the basic grips, additional grips may be added. It is assumed that if the cognitive load is too high, the user may stop using the multi-function hand or may not take full advantage of its advanced features. Measurement of cognitive load while learning to use a bionic hand will help the therapist to adjust the training pace and help the user to achieve success. An innovative, non-obtrusive method for measuring cognitive load is by tracking eye gaze. Gaze measures provide pupil diameters that indicate subjective task difficulty and mental effort. [1]

Previous studies have demonstrated that there is a connection between the need for visual feedback and learning to operate a myoelectric prosthesis [2], but few have looked at cognitive load in the learning/training process. Therapists have no objective data to help determine if a person is experiencing excessive cognitive load or when they are ready to progress to learning more advanced functions of the hand.

AIM

The aim of this project was to analyze cognitive load at various time intervals during the learning process in using a multi-function hand.

METHOD

After receiving ethics board approval and informed consent, three prosthetic users were assessed while learning to use multi-function hands. In cases where they had experience using a myoelectric hand, they were assessed using that hand as well. They went through basic skills training of learning to open and close the hand, and switch between two to three basic grips and use them to pick up and manipulate various objects. All three users had prior experience in using myoelectric control. Table 1 shows demographic information.

Table 1: Subject demographics

	Subject 1	Subject 2	Subject 3
Age	68	47	33
Level of amputation	transradial	transradial	transhumeral
Time since amputation	12 years	6 years	10 years
Previous prosthetic hand(s)	MC Pro-Control, Bebionic	iLimb Ultra (2 years of no use)	iLimb Ultra (3 years of no use)
Control of previous hand	Two site	One site	Two site (weak muscles)
Prosthetic hand assessed	iLimb Quantum	iLimb Ultra	iLimb Quantum
Control used	Two-site	Two-site	Coapt pattern recognition

Subjects wore Tobii Pro2 eye-tracking glasses before beginning initial training with the prosthetic hand. When the subject was comfortable with the use of the hand, a SHAP assessment was completed in a seated position with the table set to the appropriate height to allow the elbow to rest at 90 degrees on the table surface.

The glasses data were imported in the Tobii Pro Lab version 1.130. The data was first inspected to remove unexpected pupil changes due to sudden head movements. Then the recordings were extracted according to the activities being performed. Measurements of pupil diameter for each activity were extracted from the time when the therapist just finished her instruction and before the subjects initiated the grip until the activity was completed and the hand returned to its resting position. The normal range of pupil diameter was set at 2-4mm (during normal light) to indicate an acceptable amount of cognitive load. [3]

RESULTS

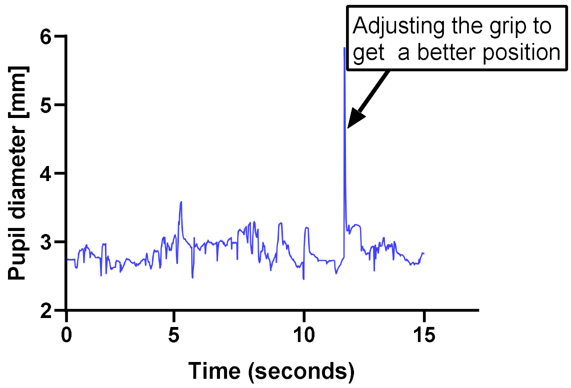
Larger pupil diameters were found in all three subjects when they used precision pinch to perform the activities "stack 4 1-inch wooden blocks" and "pick up small objects" (Table 2). Subject 3 had pupil diameters over 4mm in all the activities. From Fig.1, it shows that pupil diameters were largest when the subjects were adjusting the grip and when they had difficulties in initiating the grip.

Table 2: Pupil diameters during training

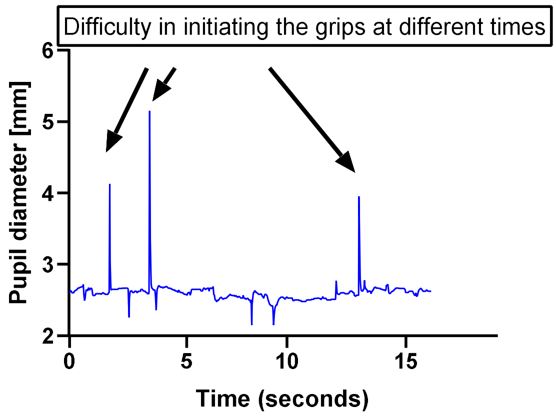
	Pupil dilation (in mm)		
	Subject 1	**Subject 2**	**Subject 3**
Pre-activity baseline (no stimuli)	R:2.52-3.21 (M=2.96, SD=0.13)	R:2.05-2.67 (M=2.45, SD=0.09)	R:2.68-3.86 (M=3.75, SD=0.15)
Activity			
Pick up ball Grip: spherical (whole hand)	R:2.42-3.63 (M=2.89, SD=0.11)	R:2.34-2.99 (M=2.56, SD=0.09)	R:3.01-**5.88** (M=4.06, SD=0.44)
Pick up drinking glass Grip: whole hand	R:2.22-3.48 (M=2.99, SD=0.20)	R:2.07-**5.17** (M=2.58, SD=0.12	R:2.36-**4.99** (M=3.88, SD=0.36)

Stack 4 1-inch wooden blocks Grip: precision pinch	R:2.39-**5.87** (M=2.95, SD=0.14)	R:2.07-**5.15** (M=2.6, SD=0.12)	R:2.61-**4.90** (M=4.20, SD=0.27)
Pick up small objects (paperclip, nail, plastic button) Grip: precision pinch	R:2.44-4.29 (M=3.00, SD=0.14)	R:1.86-**4.74** (M=2.52, SD=0.11)	R:3.05-**5.27** (M=4.32, SD=0.38)
Open plastic storage bag Grip: precision pinch	R:2.44-3.37 (M=2.86, SD=0.18)	Not performed	R:3.31-**4.37** (M=4.02, SD=0.17)
Hold playing cards Grip: lateral/key	R:2.33-3.42 (M=2.88, SD=0.17)	R:188-3.02 (M=2.48, SD=0.09)	R:2.54-**4.67** (M=3.61, SD=0.43)
Hold knife to cut playdough Grip: Lateral and between fingers	R:2.26-3.55 (M=2.90, SD=0.15)	R=2.02-2.85 (M=2.48, SD=0.13)	R:2.82-**5.83** (M=3.6, SD=0.36)
Hold fork to hold playdough Grip: lateral/key	R:2.46-**5.83** (M=2.88, SD=0.24)	R:2.02-3.07 (M=2.37, SD=0.16)	R:2.53-**4.59** (M=3.12, SD=0.30)

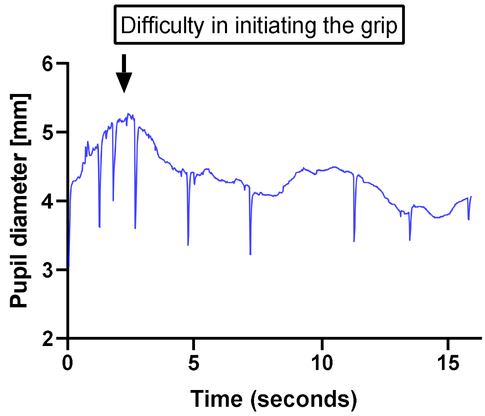
R: range, M= mean, SD =standard deviation, numbers in bold=over 4mm



Subject 1: Hold fork while cutting playdough (lateral grip)



Subject 2: Stack 4 1-inch wooden blocks (precision pinch)



Subject 3: Pick up small objects (precision pinch)

Fig.1: Changes in Pupil Diameter over time

DISCUSSION AND CONCLUSION

Based on the pupil diameters from the four grips analysed here, it seems appropriate to introduce not more than four grips during the first control training. It is unknown whether pupil diameters will decrease over time when adequate prosthetic training is given. As we can see from the results, it is cognitively demanding to learn to use a multi-function hand, especially during initiating a new grip. Further research with more prosthesis users over time and other multi-function hands is needed to confirm the study findings.

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COMPARATIVE EFFECTIVENESS AND FUNCTIONAL PERFORMANCE OF MULTIPLE DEGREE OF FREEDOM PROSTHETIC HANDS IN INDIVIDUALS WITH UNILATERAL TRANSRADIAL OR WRIST DISARTICULATION AMPUTATION

F. Clay Smither MD[1], Joseph Webster MD[1], Matthew Borgia MA[2], Sam Phillips[3], Linda Resnik PT, PhD, FAPTA[2]

[1]Hunter Holmes McGuire VA Medical Center, Richmond, VA, USA
[2]Providence VA Medical Center, Providence, RI, USA
[3]James Haley VA Medical Center, Tampa, FL, USA

ABSTRACT

The stated goals of multiple degree of freedom (DOF) prosthetic hands are to improve function and create more natural movements for the prosthetic user. This cross-sectional observational study tested 75 persons with unilateral transradial or wrist disarticulation amputation using standardized measures. Three subtypes of prostheses were compared: body-powered, myoelectric single-DOF terminal device, and myoelectric multi-DOF terminal device. In most categories there was no significant difference in performance with the multi-DOF devices. Body-powered users had better scores in two measures of dexterity compared to myo multi-DOF users. Myo single-DOF users performed better than body-powered users in one test of everyday activities.

INTRODUCTION

Multiple degree of freedom (DOF), or multi-articulating prosthetic hands are arguably the most advanced prosthetic terminal devices. The benefit of these devices includes the more lifelike hand appearance[1] and the ability to assume multiple different hand positions and grasp patterns[2] which, in theory, can enhance performance in a variety of activities. Device manufacturers also report that individual finger motion allows more natural and coordinated movements and greater precision control over delicate tasks.

However, there is limited research examining functional performance of persons using these devices. The purpose of this presentation is to compare dexterity and activity performance of users of multi-DOF myoelectric, single-DOF myoelectric and body-powered devices.

METHODS

A cross-sectional, observational study was conducted. The VA Central Institutional Review Board (IRB), Regional Command-Central IRB and the Human Research Protection Office (HRPO) reviewed and approved this study. All study participants gave voluntary informed consent.

This report is a sub analysis of a larger study of prosthesis users. Exclusion criteria included inability to wear a prosthesis for 3 hours, and any health condition that would limit participation in the study activities. The analysis presented here is limited to participants with unilateral amputation at the transradial or wrist disarticulation level.

Data was collected at one of five sites by either occupational or physical therapists. Demographics, directed history, prosthesis evaluation and physical examinations were obtained and performed. A prosthetist evaluated photographs of the prosthesis to confirm device type. Standardized measures of performance were taken, including Jebsen-Taylor Hand Function (JTHF)[3], Nine Hole Peg (NHP)[4], Box and Block[5], Southampton Hand Assessment Procedure (SHAP)[6], Activities Measure for Upper Limb Amputation (AM-ULA)[7], Brief Activities Measure for Upper Limb Amputation (BAM-ULA)[8], and Timed Measure of Activity Performance (T-MAP)[9].

Prosthesis type was classified as: body-powered, myoelectric single-DOF terminal device, and myoelectric multi-DOF terminal device. Kruskal-Wallis tests were used to compare outcomes by prosthesis type. Dunn's post-hoc tests were used to identify differences between categories of prosthesis type for all outcomes.

RESULTS

Seventy-five persons with unilateral transradial or wrist disarticulation amputation were included in this analysis. Table 1 provides demographics and prosthesis type. The participants were 97% male with mean age of 57. Trauma caused most limb loss. Table 2 describes the measures.

Kruskall-Wallis results are shown in Table 3. There were significant differences by group in JTHF small objects and heavy can items, NHP and BAM-ULA scores. Statistically significant post hoc comparisons are shown in Table 4. Users of body-powered devices had better scores of the JTHF small object tests and NHP as compared to myo multi-DOF users. BAM-ULA scores were better for myo single-DOF users as compared to body powered users.

SUMMARY & CONCLUSIONS

Despite the reported benefits of multiple degree of freedom prosthetic hands, we found no differences in fine motor or everyday activities between those using myoelectric multi-DOF terminal devices and myoelectric single-DOF devices. We did find that users of body powered prostheses had better dexterity scores on 2/10 of tests. In a test of ability to complete everyday tasks, persons using single-DOF myoelectric prostheses performed better than persons using body powered devices.

Prior studies have compared the performance of persons using body-powered and myoelectric prostheses. Hebert et al. studied a single person with transhumeral amputation performing a box and blocks test with a body-powered prosthesis, then 13 months after targeted muscle reinnervation and training with a myoelectric prosthesis. He was able to move 49 blocks with a body-powered prosthesis but only 20 blocks with the myoelectric prosthesis. Motion analysis showed less compensatory trunk movements with the myoelectric device and more natural elbow movement.[10]

Meredith compared the Ottobock Electric Hand, Ottobock Griefer, Hosmer Senergetic Prehensor and body-powered hook in NHP, Box and Blocks and JTHF tests. They evaluated three subjects with transradial amputations, two of whom used a body-powered hook daily and one who used a myoelectric hand. The subjects were trained with Greifer and Synergetic Prehensor prior to testing. In NHP, all three were fastest with Synergetic Prehensor. In the other two tests, the fastest times were distributed between the different devices.[11]

When considering why persons using body-powered prostheses performed better on the NHP and JTHF small items, it may be that multi-DOF terminal devices are complex to use and thus slower to control in fine motor movements, particularly given the need to change grasp patterns and to select the most appropriate grasp for specific tasks.

Our study found that persons using myoelectric single-DOF prostheses had higher scores than body powered users on the BAM-ULA, indicating that they were able to complete more activities as compared to body powered users. Given our findings, we compared scores of individual tasks of the BAM-ULA using Fisher's exact tests to determine if there were specific items that were driving BAM-ULA sub-group differences. We found that scores differed in two items: remove cap from water bottle and drink and lift one-gallon jug. It is likely that body powered users had difficulty regulating grip force in grasping the water bottle, and that they lacked the grip strength and/or could not position their terminal devices to lift the one-gallon jug.

These findings should be considered preliminary due to small sample sizes for groups. Additionally, we did not control for training or years of experience. Subjects were tested using their own prostheses, and some of the tasks tested were not activities that the users routinely performed with their prosthesis (such as brushing hair). Future study involving larger sample sizes are needed to confirm or refute these finding and to evaluate differences by prosthesis make and model.

Table 1: Demographics and Prosthesis Characteristics of Participants

	Body powered (N=45)	**Myo single-DOF (N=12)**	**Myo multi-DOF (N=18)**	**All (N=75)**
	Mn (sd)	**Mn (sd)**	**Mn (sd)**	**Mn (sd)**
Age	62.8 (16.2)	45.8 (16.1)	48.4 (14.3)	56.6 (17.3)
Years since amputation	30.9 (20.5)	14.8 (12.9)	16.5 (15.9)	24.2 (19.7)
	N (%)	**N (%)**	**N (%)**	**N (%)**
Gender				
Male	45 (100.0)	11 (91.7)	17 (94.4)	73 (97.3)
Female	0 (0.0)	1 (8.3)	1 (5.6)	2 (2.7)
Etiology of amputation*^				
Congenital	2 (12.5)	0 (0.0)	1 (16.7)	3 (13.0)
Combat	20 (51.3)	4 (50.0)	3 (21.4)	27 (44.3)
Accident	16 (41.0)	6 (75.0)	8 (57.1)	30 (49.2)
Burn	2 (5.1)	1 (12.5)	1 (7.1)	4 (6.6)
Cancer	2 (5.1)	0 (0.0)	2 (14.3)	4 (6.6)
Diabetes	1 (2.6)	0 (0.0)	0 (0.0)	1 (1.6)
Infection	7 (18.0)	0 (0.0)	1 (7.1)	8 (13.1)

*Etiology of amputation: respondents could indicate multiple etiologies
^ Etiology of amputation was not collected for all participants

Table 2: Description of Performance Measures

	Construct	Item Content	Rating Criteria	Interpretation
Jebsen-Taylor Hand Function (JTHF)	Dexterity	7 separate tests of fine motor activities: writing, page turning, small objects, eating, placing checkers, light cans, heavy cans	Performance speed; items / per second (modified scoring)	Higher scores are better
Nine Hole Peg	Dexterity	Accurately place and remove 9 plastic pegs into a pegboard	Timed Measure; item/s second (modified scoring)	Higher scores are better
Box and Block	Dexterity	Number of wooden blocks transported in 60 seconds	Performance speed; Total number of blocks transported	Higher scores are better
Southampton Hand Assessment Procedure (SHAP)	Dexterity/ Index of Function	26 unilateral timed tasks of hand function: 12 abstract tasks and 14 activities of daily (such as zipping, pouring, buttoning).	Performance speed	Higher scores are better
AM-ULA	Activity performance	18-everyday tasks: brush/comb hair, don t-shirt, doff t-shirt, button shirt, zip jacket, don socks, tie shoes, drink from a cup, use fork, use spoon, pour 12 oz can, write, use scissors, turn doorknob, hammer nail, fold towel, use phone, reach overhead	Each item is rated on: task completion: speed, movement quality, skillfulness of prosthesis use and independence.	Higher scores are better
BAM-ULA	Activity performance	10 everyday tasks: tuck in shirt, lift 20 lbs, open and drink from water bottle, remove wallet from back pocket, replace wallet, lift gallon jug, open and pour jug, brush/comb hair, use a fork, open door with round knob	Ability to complete each task (yes/no). Total score is the number of completed activities	Higher scores are better
T-MAP	Activity performance	5 everyday activities: drink from a cup, wash face, food preparation, eating, dressing	Timed Measure: sum of time to complete each activity	Lower scores are better

Table 3: Functional Outcomes by Device Type

	Body powered (N=45)	**Myo single-DOF (N=12)**	**Myo multi-DOF (N=18)**	**Kruskal Wallis**
	Mn (sd)	**Mn (sd)**	**Mn (sd)**	**p**
Dexterity				
JTHF				
Writing	0.49 (0.30)	0.41 (0.26)	0.52 (0.30)	0.4274
Page turning	0.13 (0.09)	0.14 (0.10)	0.12 (0.07)	0.8182
Small objects	0.11 (0.07)	0.11 (0.11)	0.07 (0.09)	**0.0288**
Eating	0.18 (0.12)	0.17 (0.14)	0.14 (0.09)	0.4160
Checkers	0.08 (0.06)	0.08 (0.09)	0.12 (0.08)	0.0957
Light cans	0.20 (0.13)	0.22 (0.11)	0.28 (0.15)	0.2295
Heavy cans	0.20 (0.17)	0.26 (0.12)	0.25 (0.14)	**0.0481**
Box and Blocks	19.00 (8.73)	14.27 (7.88)	15.28 (6.19)	0.0645
Nine Hole Peg	0.06 (0.05)	0.06 (0.06)	0.01 (0.01)	**0.0008**
SHAP IOF	42.4 (18.4)	39.3 (23.1)	40.2 (15.0)	0.8828
Activity Measures				
AM-ULA	14.9 (5.3)	14.9 (7.7)	16.4 (6.5)	0.5800
BAM-ULA	6.6 (2.1)	9.2 (1.0)	8.0 (1.6)	**0.0023**
T-MAP (mins)	5.0 (1.8)	3.9 (0.6)	3.9 (0.9)	0.0810

Table 4. Statistically Significant Group Differences: Results of Dunn's Test

	Body powered vs. myo single-DOF	**Body powered vs. myo multi-DOF**	**Myo single-DOF vs multi-DOF**
JTHF Small objects	No difference	Body powered is better	No difference
JTHF Heavy cans	No difference	No difference	No difference
Nine Hole Peg	No difference	Body powered is better	No difference
BAM-ULA	Myo single-DOF is better	No difference	No difference

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THE COMPARISON OF FUNCTION AND USEFULNESS OF VOLUNTARY CLOSING AND VOLUNTARY OPENING BODY-POWERED PROSTHESES

Chika Nishizaka1), Sayaka Fujiwara1), Satoshi Endo1), Hiroshi Mano1,2), Kengo Ohnishi3), Nobuhiko Haga1)

1)Department of Rehabilitation Medicine, The University of Tokyo Hospital

2)Department of Rehabilitation Medicine, Shizuoka Children's Hospital

3) Tokyo Denki University

ABSTRACT

A comparison of functionality of the voluntary opening (VO) terminal device and the voluntary closing (VC) terminal device was performed using body-powered prostheses simulator on 52 non-amputee adults. We compared The Southampton Hand Assessment Procedure (SHAP) score and time required to complete SHAP's 26 tasks. The results show that the VC terminal device is easier to operate than the VO terminal device, when strong gripping force and quick reaction time is required.

BACKGROUND

Voluntary opening (VO) terminal device is often selected for body-powered prosthesis. Yet, there are tasks where voluntary closing (VC) terminal device are known to be more useable by experience. There are studies comparing the pinch force and cable efficiency of the hands and hooks of VO/VC terminal deices[1.2)]. However, these research do not compare the actual ease of use of the VO and VC terminal device in daily living activities. There is also a report which discusses the VOC type, that is capable of switching between VO and VC, is most efficient[3)], although the actual situation is not clear and further research is need. We believe the necessity to distinguish the characteristics of the VO and VC for selecting the best terminal device for the amputee's daily activities when prescribing body-powered prosthesis. This research evaluates and compares the VO and VC terminal device with a body-powered prosthesis simulator donned on the left arm by non-amputee subjects.

OBJECT AND METHOD

This study was approved by the Ethics Committee of the Faculty of Medicine of the University of Tokyo. The participants were non-amputee adults (n=52, 26 males and 26 females), average age was 30.6 years, all right-handed (Edinburgh dominant hand test was more than 50 points). Written consent was obtained from all participants. The Southampton Hand Assessment Procedure (SHAP) was performed with VOC terminal device (infinite Equilux) attached to the body-powered prosthesis simulator, and with the participants' left sound hand. SHAP is an upper limb function test developed in the UK, in 2002. It consists of 2 tests: a12-item test for daily movements with different shapes (e.g. spheres and cylinders) and weights, and a 14-item two-handed motion test. SHAP is an assessment tool that evaluates the time required to perform each task and automatically calculates the score using its own formula by inputting it into the SHAP Web program. In addition to the total score, scores for each of the six hand movement patterns (Tip, Lateral, Tripod, Spherical, Power, and Extension) are calculated as functionality profiles. The lowest possible score is 0, while the highest is 100 + α, and the cut-off score of a normal person is 95.

All subjects had no experience using body-powered prosthesis simulator or SHAP. The subjects were divided into two groups: the VO group which performed the test with the VO terminal device and the VC group with the VC terminal device. The two groups were randomized by stratification equally so that there was no difference in the number, age, and gender. The SHAP scores and the time required to perform each task was compared between the two groups. In addition, cable efficiencies were measured when using the body-powered prosthesis simulator with each setting. The SHAP scores, task perform time, and Edinburgh dominant test score were compared by Wilcoxon signed rank test. Statistical analysis was performed using JMP® Pro 14.2.0 (SAS Institute Japan) and $p < 0.05$ was considered significant.

RESULT

VO group was 26 people, 13 males and 13 females (average age: 30.3 years, average Edinburgh dominant test points: 97.1) and VC group was 26 people, 13 males and 13 females (average age: 30.9 years, average Edinburgh dominant test points: 95.2). There were no differences between the two groups in average score of the Edinburgh dominant hand test ($p=0.23$).

The cable efficiency was 48.8% for the VO and 50.6% for the VC. The average score of SHAP with the participants' left hand was 97.2 points (ranging from 93 to 103 points), and there was no difference between the two groups (VO group average 97.3 points, VC group average 97.2 points | $p=0.75$). Table 1 shows the total score of SHAP and the scores of the six hand movement patterns. There was no significant difference between the two groups in the SHAP total score ($p=0.20$). However, for the extension, the score was significantly higher in the VC group: average of 36.8 points in the VO group and average of 42.9 points in the VC group ($p=0.005$). For all six movement patterns, the VO group's average score did not exceed that of the VC group.

Table 2 shows the time required to perform each SHAP task. When comparing the time, the VC group was significantly faster than the VO group in the three tasks: Heavy Power (VO group 9.7 seconds, VC group 8.4 seconds, $p=0.049$), Heavy Extension (VO group 8.9 seconds, VC group 7.1 seconds | $p=0.04$), and Cutting (VO group 158.9 seconds, VC group 89.6 seconds | $p<0.005$). In addition, in 19 of the 26 tasks, the average time required to perform the task were shorter in the VC group than of the VO group, including those with no significant difference.

DISCUSSION

The SHAP task with Power and Extension task is conducted handling light and heavy object of the same shape. In both Power and Extension task, there was no significant difference in the handling light objects. However there was significant difference in handling heavy objects, and was faster in the VC group. The cutting task involves holding a knife in the terminal device and pressing into the clay, which also requires a strong grip. The significant difference between the groups suggests that the VC type body-powered prosthesis is particularly useful for tasks required to generate high grip strength.

Regarding the time required to accomplish each task, the VC type was faster to grasp the object. The initial opening movement of the hook before grasping the object in the VO type makes it slower to accomplish the task.

The experimental results of VC type with higher score and shorter time than VO type for tasks that require strong gripping force or that require quick operation indicate that VC terminal device can be prescribed when the amputee is focused on these tasks in daily life, recreation, and occupation.

There are limitations to this study. The participants of this study are non-amputees, and it was the first time for all the participants to operate the body-powered prosthesis. Future studies should make efforts to measure the amputees who use the body-powered prostheses for daily use.

CONCLUSION

The aim of this study was to understand which prosthetic tool is more appropriate based on movement types and the needs of patients and prescribe a prosthesis that is easier to use the VC type moves significantly faster according to patient’s lifestyle.

Table 1 : SHAP scores

	Six patterns Score						Total Score
	Sperical	Tripod	Power	Lateral	Tip	Extension	
VC	43.92	23.84	37.5	47.34	31.84	49.07	46.53
VO	40.57	20.96	35.11	44.03	26.84	36.84	42
p	0.26	0.23	0.29	0.49	0.29	0.0051*	0.2

*：p<0.05 **：p<0.005

Table 2 : Time spend on each SHAP task

Task	Abstract Object - Lightweight						Abstract Object - Heavyweight					
	Spherical	Tripod	Power	Lateral	Tip	Extension	Spherical	Tripod	Power	Lateral	Tip	Extension
VC	6.25	12.2	5.78	9.47	11.77	7.3	12.47	7.3	8.47	9.02	7.65	7.18
VO	7.61	9.3	6.39	10.74	13.68	19	9.38	9.2	9.71	8.99	13.15	8.95
p	0.14	0.99	0.48	0.8	0.43	0.059	0.66	0.29	0.049*	0.99	0.35	0.041*

Task	Activities of Daily Living													
	Coins	Button Board	Cutting	Page Tuning	Jar Lid	Jug Pouring	Carton Pouring	Full Jar	Empty Tin	Tray Lift	Key	Zip	Screwdriver	Door Handle
VC	46.35	38.57	89.62	10.88	18.51	24.61	23.51	10.07	7.58	7.96	4.81	16.32	22.51	5.31
VO	51.86	45.22	158.99	11.99	17.89	31.79	24.71	59.97	8	8.67	5.04	11.32	18.11	4.98
p	0.11	0.18	0.0002**	0.1	0.85	0.19	0.21	0.42	0.6	0.27	0.26	0.63	0.54	0.46

*：p<0.05 **：p<0.005

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DATA LOGGING DURING PATTERN RECOGNITION CALIBRATION AS A REMOTE DIAGNOSTIC TOOL

Laura A. Miller, PhD, CP,[1,2] Kristi L. Turner, DHSc,[1] OTR/L Ann M. Simon, PhD[1,2]

[1] Center for Bionic Medicine, Shirley Ryan AbilityLab, Chicago, IL, USA
[2] Northwestern University, Chicago, IL, USA

ABSTRACT

Pattern recognition control uses EMG from the entire residual limb to more intuitively control prosthetic devices. However, this requires a more intimate socket fit to maintain contact with these additional sensors. When users complain of issues with control, it can be difficult to diagnose if the issue is a need for additional practice and training or if there are issues related to the prosthetic fit that need to be addressed. Pattern recognition does allow the recalibration of the system by the user in any location. By analysing the data logging of calibration data in a pattern recognition system, it is possible to better identify the cause and potential solution in a remote setting.

INTRODUCTION

With pattern recognition (PR), multiple EMG channels can be used as input with all of the information used 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 [1]. The user needs to show the system each movement (calibrate the controller), which can be done by following prompts on a computer interface or following along with the prosthesis while it is moved through the different available movements. EMG is recorded by the controller and the classifier is then calculated.

For this PR to be successful, the EMG channels must maintain good contact with the residual limb. When fitting a user in the office or a therapy environment, the EMG quality can be monitored as the user begins to perform functional tasks in different planes of movement and adjustment to fit made as needed. However, different environments temperatures and weight gain/loss can all affect signal quality.

When the user lives nearby it can be easy to have them come in for regular rechecks and adjustments; however, when a user lives far away, it can be difficult to troubleshoot the issue and identify if the issue with control is related to EMG quality or if the issue might be related to the need for additional training and/or a review of the patterns of movement associated each degree-of-freedom.

As part of a study related to pattern recognition control of a transradial prosthetic system, users from across the country were recruited for home trials. During the home trial subjects were instructed to send home logs each week. However, there were instances of poor control noted and it was not logistically possible to bring in subjects for return rechecks. Since, during pattern recognition calibration EMG data are recorded and used to create the classifier, this property of the controller was used to collect data that could be used in a diagnostic manner for evaluation of fit and function. A protocol was developed to record information in various positions to allow repairs and adjustment to take place without an in person visit. This technique was also used to verify fit prior to beginning home trials.

METHODS

Eight individuals with a unilateral transradial amputation were fit with a Coapt pattern recognition system [2] passive wrist, and i-limb TMR revolution [3]. The study (including the ability to collect and record EMG data) was approved by the Northwestern University IRB. During the calibration process of pattern recognition control, data were recorded to be used to generate a classifier as the prosthesis moved through the various movements. The system would first collect EMG of the users' arm at rest (to align with "no movement" of the prosthesis). The prosthesis would then cycle through all of the enabled grasp patterns, opening and closing of each grasp 2 times. For this study, all calibration data was recorded and stored on the embedded controller for later post-processing.

Users were provided OT prior to participating in an 8-week home trial to evaluate their pattern recognition control of the multiarticulating hand. They were trained to calibrate their prosthesis whenever they felt their control had degraded. They checked in weekly using a home log system. Logged issues or calls to the prosthetist/OT over this 8-week window often needed to be followed up and these issues were often difficult to diagnose. In a clinical setting, users would be brought in for a recheck to evaluate fit and function. Since this was not always possible due to distance, alternative options were explored.

Since EMG was recorded for later evaluation during the calibration, a fitting evaluation protocol was designed to use this recording for diagnostics. All subjects had a minimum of 3 grasp patterns enabled. This allowed for the collection of 13 (no movement plus 4 cycles * 3 grasp patterns) 3-second data blocks. Users were prompted to perform specific movements in various positions during the data recording phases of calibration. The order of movements requested was recorded so that the data collected could be mapped to arm position/contraction type. Table 1 shows the protocol developed and used in most cases. For these diagnostic trials, when collecting movement and maximum voluntary contraction (MVC) data, subjects were instructed to move the arm around in space when the device was moving. When conducted remotely, this prompting occurred via phone call/skype to assist with timing. Six participants used the evaluation protocol developed to diagnose fit and training issues. Some subjects also performed the protocol in lab as a "check out" of fit prior to starting the home trials.

Table 1: List of prompted movements for evaluation of EMG quality

Arm supported: Regular calibration with the arm supported (resting on a table)
*Arm down at side: R*egular calibration with the arm relaxed down at the side (hanging)
Arm in front of body: Regular calibration with the arm in front (as if shaking hands)
Arm sweeps and MVC (Maximum Voluntary Contractions) During the data collection blocks, the subject was prompted as follows: 1. Arm down at side and contract all forearm muscles at MVC 2. Arm in front and contract all forearm muscles at MVC 3. Arm out to side and contract all forearm muscles at MVC 4. Forearm relaxed and sweep arm from down at side to up to cabinet level and back down, diagonally 5. Forearm relaxed and sweep arm side to side at cabinet level 6. Forearm relaxed and push in on socket and wiggle 7. Forearm relaxed and pull slightly on socket
Subject prompted to doff and re-don system and repeat the following:
Arm in front of body
Arm down at side

Data were downloaded from the embedded controller for further processing. In most cases this occurred when the arm was sent back by mail (cheaper than flying the user back for an in-person visit) or by downloading to a study computer sent to them. A custom Matlab script was written to import the files and create graphs of the 8 channels of EMG. Data were plotted with each movement concatenated in order (i.e., no movement followed by open/close/open/close of each configured grip) with the channels shown 1-8 from top to bottom. The date/timestamp of the data was included in the title for reference and custom titles could be applied. Some of the issues (mechanical and therapy related) that were possible to diagnose:

- No issues with EMG (i.e., clean) EMG during normal use but intermittent EMG saturation either at different positions or during MVCs: Electrode lift off from contraction or position. Or an intermittent loose wire
- Constant EMG saturation or noise: Broken wire or loose wire
- EMG saturation during muscle contractions: EMG gain too high or user contracting too hard No EMG noted at all (flatline): broken wire or electrode shorted
- Clean EMG collected but hand did not move properly during calibration: hand requires repairs
- EMG improperly timed contractions of regular training (contraction only in small part of each window): subject needs more training
- EMG barely detectible for all movements: EMG gain too low, EMG location not ideal, or contractions too light
- Clean EMG but user has poor control after recalibration: user needs more training/alternative imaging for different grasp patterns
- EMG after redonning very different than first 2 trials: user needs more practice with repeating proper donning or recreating grasp patterns

RESULTS

The protocol was used throughout the study to confirm socket fit and EMG quality when subjects were in the lab for testing/fitting and also when subjects experienced control issues at home. Figure 1 shows an example of early fitting with the pattern recognition system. The subject had reasonable control; however, upon reviewing the calibration data it was noted that the EMG on Channels 2, 4, 5 and 6 was significant smaller than that of the other channels for all movement classes; therefore, the gains were increased prior to beginning the trial.

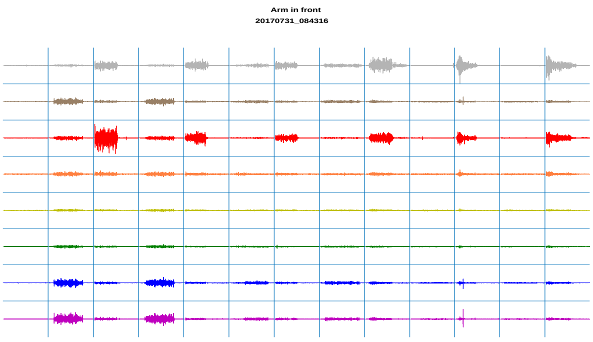


Figure 1: Gain imbalance: EMG gain on Channels 2, 3, 4, and 6 were subsequently increased

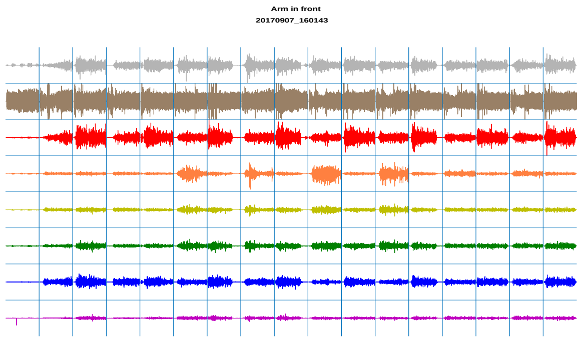
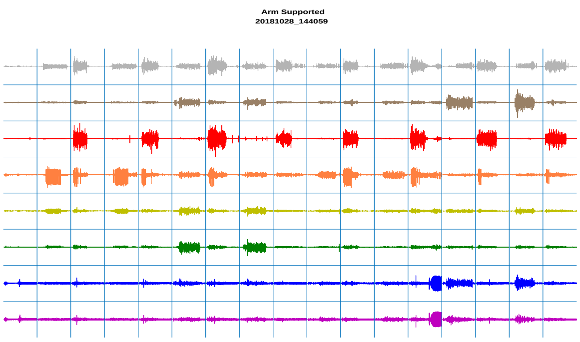


Figure 2: EMG analysis after arm sent in for adjustment. Noise seen on Channel 2 and loose wire located inside socket



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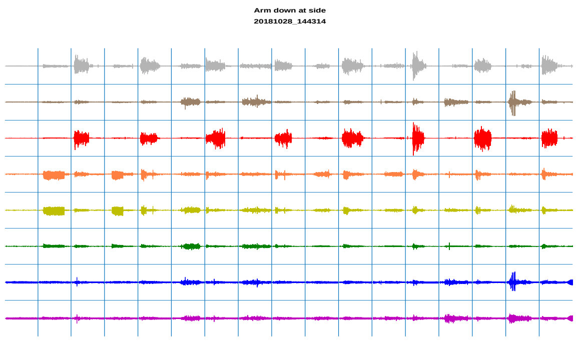

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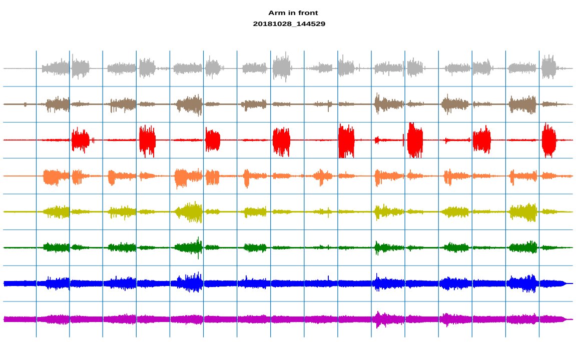

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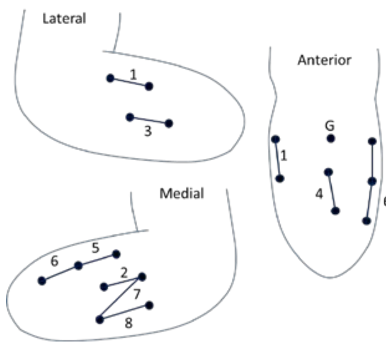

d

Figure 3: Remote troubleshooting with one subject. The 4 images show the data collection for a) arm resting, b) arm at side, c) arm in front, and d) channel locations in the socket. The 8 EMG channels are shown 1-8 from top to bottom in a-c. The thin vertical lines delineate where the EMG from the various movements (4 different hand grasp patterns) have been concatenated. Each vertical band represents 3seconds of data.

Figure 2 shows one example of the evaluation protocol used for remote troubleshooting. The subject had complained of poor control and was prompted through the diagnostic protocol prior to sending his arm in for review. Upon inspection of the data, channel 2 showed consistent noise across all movements and positions. This EMG contact was assessed and it was found that the wire connection inside the socket at the ring terminal to the EMG dome had broken during use. Though this failure would likely have been found with a thorough inspection of the device, the evaluation protocol made diagnosis and repair much quicker.

A more complex example can be found in Figure 3. This subject had previously undergone a revision surgery and was experiencing continued volume loss during the home trial. It was identified during planned follow up that he was having issues with control in some positions. The EMG from the evaluation protocol was compared to the locations of the electrode channels within the socket. The 4 images show the data collection for a) arm resting, b) arm at side, c) arm in front, and d) channel locations in the socket. When the arm was resting, it appeared that the soft-tissue was pulling away from the anterior channel (channel 4) and then pulling away from the posterior channels when the arm was extended (channels 3, 7, 8). Spacers were added to increase the depth of compression of the electrode domes on these 4 channels and the prosthesis was returned to the user. He reported improved control after return of the device and the EMG quality was verified at his next scheduled in person visit.

Other cases were noted where, upon completion of the evaluation protocol, the EMG quality was good. In these cases, the subjects would continue to work with the Occupational Therapist either in person or remotely to identify phantom movements that would create EMG unique to each grasp pattern.

DISCUSSION

Pattern recognition control has become more common in upper limb prosthetic fittings; however, the increase number of EMG channels associated with these systems can make troubleshooting fit and function difficult. It is possible to visually review the EMG when the user is present but if issues arise a way of assessing the issue remotely is useful.

When EMG calibration data is recorded onto the prosthesis, this feature can be used to collect data to assess EMG and fit. This protocol was used on six individuals participating in home trials and was useful to diagnose loss of contact and broken wires, which were repairable without an in-person visit. In this study we needed to ship the prosthesis back to physically collect the data from the arm (or ship a laptop to the user), but if the data were downloaded remotely to a secure server it would be possible to identify problems with training or other issues that don't require repair to be completely resolved remotely. Additionally, the ability to remotely download the data would have allowed subjects to repeat the series of diagnostic training sessions to confirm that the repairs/socket modifications resolved the issue.

This evaluation protocol was also useful for confirming fit prior to the home trial by prompting the user to control the device in various planes of movement and as a baseline before home trial in case issues would arise later. This paper presents work done for a research study, but a similar evaluation protocol would be useful in the clinical environment to assist the prosthetist and occupational therapist to determine when it is necessary for a user to schedule follow up care.

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DEMOGRAPHIC DIFFERENCES IN THE UPPER LIMB PROSTHETIC REHABILITATION EXPERIENCE

Daire McGuinness[1], BSc, Diane Atkins[2], OTR/L, FISPO

[1]*Össur, R&D, Livingston, UK,* [2]*Baylor College of Medicine, Houston, TX*

ABSTRACT

This study aimed to evaluate trends in the prosthesis provision and training experience of individuals with upper limb absence and whether these trends were associated with any demographic factor. Furthermore, we evaluated whether the rehabilitation experience was associated with quality of life, health markers and other measures of rehabilitation success. Results of this study indicate demographic differences in upper limb prosthetic rehabilitation as well as trends in the effect of the prosthetic rehabilitation experience on patient outcomes.

INTRODUCTION

The loss of one hand can significantly affect the level of autonomy and the capability of performing daily living, working and social activities. [1] Degree of independence is one of the three indicators of Functioning, Disability and Health in the WHO International Classification [2] with maintenance of independence in activities of daily life being a key objective of post-amputation occupational therapy. [3] While determining the parameters which demonstrate "successful use" of an upper limb prosthesis is a complex topic, considering the myriad functions of the intact hand and the highly individual goals of potential users, [4] [5] degree of independence is a parameter in many functional performance measures. [6] This study aimed to identify demographic trends in individuals with upper limb absence associated with prosthesis use, rehabilitation and daily life. The results presented here indicate a strong association between gender and the prosthetic rehabilitation experience.

METHODS

Subjects

The study was recruited via email to the Amputee Coalition members database and displayed on the Amputee Coalition social media platforms. It is therefore assumed that the responses are majority North American in origin although respondent location or origin information was not recorded. Eligible participants were individuals over the age of 18 with unilateral or bilateral, acquired or congenital upper limb absence at any level. Subjects were eligible to participate in the study only once. Of a total n=309 individual responses, n=9 subjects did not complete the eligibility questions and were therefore not enrolled in the study. A further n=9 who were eligible to participate did not complete the study and were withdrawn due to incompleteness of the responses. A total of n=292 responses were included in the analysis.

Data Collection and Analysis

The study was a non-interventional, retrospective, cross-section design conducted with the approval of the NEIRB (#:120190122) consisting of a self-drafted online questionnaire and two validated outcome measures; Quick-Disability of the Shoulder Arm and Hand (QuickDASH), [7] and the EuroQol standardised measure of health status (EQ-5D-5L) [8]. Questions were grouped into categories as follows: personal demographics; prosthesis fitting and training history; current prosthesis use, activities and satisfaction; employment and activity trends. To evaluate differences in proportions, Pearson's Chi-squared significance test or the 2-sample significance test for equality of proportions were applied at a significance level alpha=0.05. Whenever needed, a continuity correction was applied for better approximations. All statistical analyses were conducted using R (version 3.6.2) software**.** [9]

RESULTS

Gender Demographics

A notable result of the study is the gender balance of respondents. It is generally accepted that the upper limb absence population trends to a male majority, [10] with females estimated to make up 20-30% of the total population. [10] [11] Conversely, in our study, female respondents were in the majority at 50.17% of the total population (46.49% male, 1.67% transgender or non-binary, 1.67% preferring not to answer). Acquired limb loss is understood to be more prevalent amongst males than females; [10] however, the prevalence of congenital limb deficiency (in the US) appears to be relatively equally distributed. [12]

In our study, 37.78% (n=57) of female respondents indicated their limb absence was congenital. Conversely only 11.85% (n=16) of male respondents indicated their limb absence was congenital. Although notable, this difference was not found to be statistically significant (p=0.06493). Congenital limb absence was indicated by 24.83% of the total respondent population.

Golden Period/First Fitting

The “golden period” in prosthetic rehabilitation is the concept that the earlier the prosthesis can be provided to a patient to use in training and therapy, the higher will be the rate of acceptance of the device and likelihood that the patient will become adept at using it as a helpful tool, [13] or be a “successful user”. [5] The “golden period” is understood to be within 30 days of amputation [5] and was first introduced by Malone et al, 1984. [14] Despite this, it is known that achieving prosthetic fitment within 30 days of an upper limb amputation is challenging and is not achieved in many cases.

This was reflected in our study, in which only n=13 (4.51%) of respondents had their first prosthetic fitting within 30 days of their amputation. The most common duration between amputation and first prosthetic fitting was indicated to be ~6 months (n=59, 20.49%). In our study, those who indicated they were currently using a prosthesis were more likely to have had their prosthetic fitting at a time within six months of amputation (p=0.0008457).

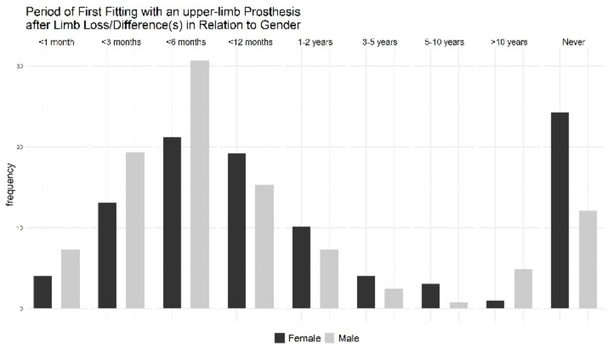

Figure 1: Time of First Fitting in relation to Gender

Females (n=38, 26.21%) were significantly less likely than males (n=71, 52.98%) to have received their first prosthetic fitting at a time within six months of amputation (p=4.646e-06). In fact, a greater frequency of females (n=24, 16.55%) than males (n=15, 11.1%) reported they had never been fit with a prosthesis, although a statistically significant difference (p=0.1973) was not found.

Reasons for Delay

Adjusting for those who perceived no delay in their prosthesis fitting (n=115. 40.49%), wound healing (n=72, 25.35%) and insurance coverage issues (n=64, 22.54%) were the most frequently indicated factors which had contributed to delay a prosthesis fitting. Interestingly, “no perceived delay” (40.49%) does not correlate with delay as reported by fitting period, if delay is considered as any fitting out-with the “golden window” (4.51%). Males were significantly more likely to report that “Physical readiness” (p=0.033) and “Wound healing” (p<0.001) caused a delay in their prosthetic fitting than females. Females were more likely to have their prosthetic fitting delayed by therapist availability issues than males (p=0.013).

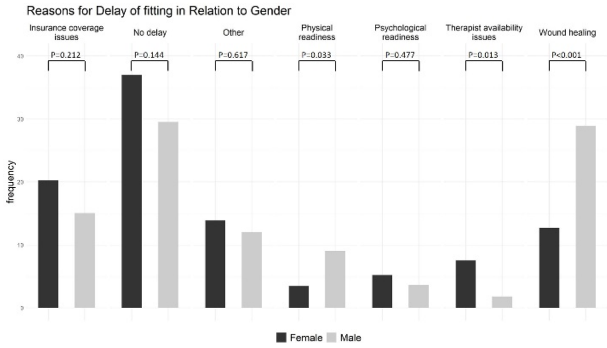

Figure 2: Reasons for fitting delay in relation to Gender

Training Received

In a systematic review, most included papers agreed that rehabilitation is vital to functional integration of upper-limb prostheses. [15] Despite the widespread agreement in the field there is a disparity between prosthesis provision and training. In our study only n=41 (14.24%) respondents reported they had never been fit with a prosthesis. However, n=102 (35.42%) of the total population reported they had never received training to use an upper-limb prosthesis, at a similar frequency to that reported by Ostlie et. al., 2012; 30.6% [16] and 31.1% [17]. In our study, those who had received prosthetic training were more likely to be currently using a prosthesis than those who had received no prosthetic training (p=4.053e-08).

Prosthesis Use

In our study, n=167 (58.80%) of respondents indicated they were currently using an upper limb prosthesis. A total of n=117, 41.20%, respondents indicated they were not currently using an upper limb prosthesis. In our study, although the frequency of males currently using any prosthetic device (n=85, 65.39%) was greater than the frequency of females currently using any prosthetic device, (n=76, 53.15%) this was not found to be statistically significant (p= 0.058).

There was a statistically significant difference in the types of prostheses currently used by male and females (p=0.000473). Evaluation of Pearson’s standardised residuals indicate that the body-powered and passive functional prosthesis types had most influence on differences in gender. Body-powered prostheses are understood to be the most prevalently used type of device in the US. [18] It is believed that females have different requirements over their prostheses than males.[19] One study showed females to be more likely to use cosmetic devices and less likely to be users of actuated devices, as compared to males. [19]

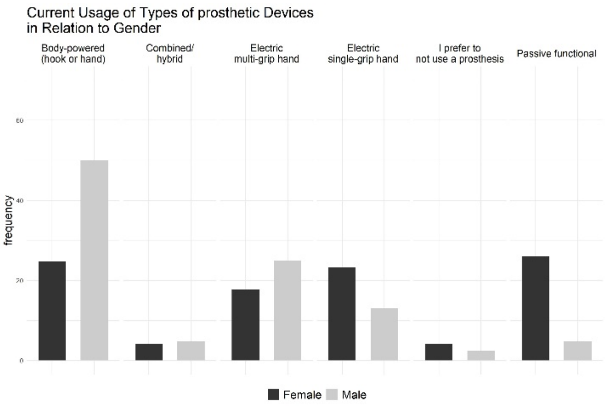

Figure 3: Current usage of types of prosthetic device used in relation to gender

Our study showed that the body-powered device was the most frequently used type of device by male respondents (n=42, 50.00%). In comparison female respondents used all device types relatively equally, body-powered (n=18, 24.66%), electric multi-grip and single grip combined (n=30, 41.10%); passive prostheses (n=19, 26.03%). Females used passive prostheses (n=19, 26.03%) at a greater frequency than males (n=4, 4.76%). Electric multi-grip and single-grip devices were used at an approximately equal frequency by both groups; females (n=30, 41.10%); males (n=32, 38.10%).

Differences in the rate of prosthesis use between males and females may be explained by a difference in the types of activities the prosthesis is required to be used for. This was not reflected in our study, in which there were no significant differences between males and females in terms of activities the prosthesis is used for.

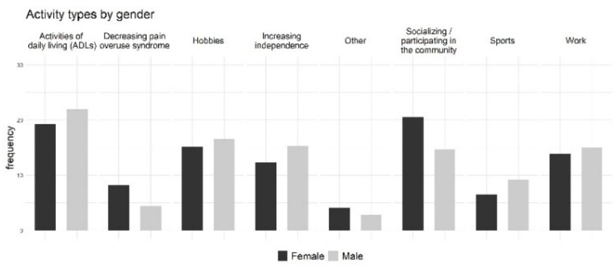

Figure 4: Activities prosthesis is used for in relation to Gender

Prosthesis Non-Use

Of the n=117 respondents who indicated they did not currently use a prosthesis, body-powered prostheses were the most frequently rejected type of device over-all (n=45, 38.46%) with electric multi-grip hands the least frequently rejected (n=14, 11.97%) over-all. There were no significant differences found in rejection rates by gender, which appear to approximately follow prescription rates.

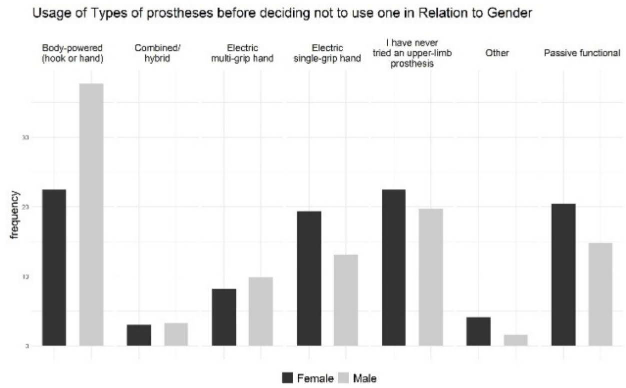

Figure 5: Usage of types of prostheses before deciding not to use one in relation to gender

Reasons for Non-Use of Prostheses

Some papers suggest that the higher rejection rate relates to a predisposition towards the aesthetics of the prosthesis in the female population, [12] inferring that prostheses do not provide aesthetic needs in females. In a further study, Biddiss & Chau reported that the type of prosthesis fitted (i.e. body-powered or myoelectric) did not appear to affect long-term use, but that passive devices were associated with higher rejection rates, [20] suggesting that insufficient functionality is also a key factor in cases of rejection.

In our study, reasons for not currently using a prosthesis were reported equally between genders in nine out of ten parameters. A significant difference was found for only one indicator, in that males were more likely than females (p= 0.014) to indicate they did not use a prosthesis because of insurance coverage issues.

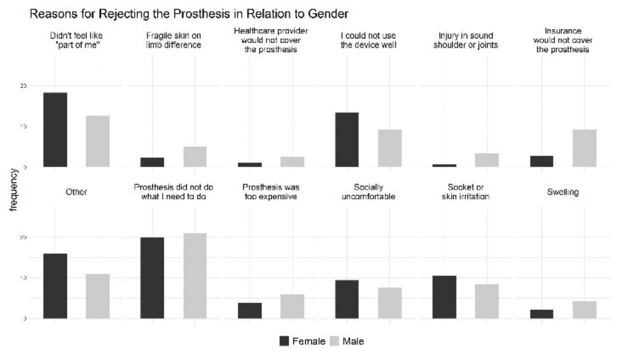

Figure 6: Reasons for rejecting prostheses in relation to Gender

In our study, the most frequently indicated reason for currently not using a prosthesis was functional. The reason "Prosthesis did not do what I need to do" was indicated by n=62 (52.99%) of our population not currently using prostheses.

DISCUSSION

The results of this study show that only 4.58% of respondents received a prosthesis within the "golden period" of 30 days from the time of amputation. Our study suggests that fitting within 6 months equates to a "better outcome" or greater likelihood of current prosthesis use, which supports current rehabilitation practices. Known challenges in the early fitting process were well represented in our study, with wound healing and insurance coverage issues being the most frequently reported. Interestingly, the most common response to this question was that "no delay to fitting" was perceived by the individual, which may be a result of expectation management by experienced clinical teams.

Our study revealed a statistically significant likelihood for those who had received prosthesis training to be currently using a prosthesis. This finding further cements the link between a thorough rehabilitation and training programme and a "better outcome" or greater likelihood of current prosthesis use. Further research is indicated to understand and alleviate specific barriers to fitting and training access.

Significant differences between genders were reported in the time to first fitting as well as perceived causes of fitting delay, however these barriers to treatment did not correlate to a significant difference in use of prostheses in daily life. These gender-associated differences in rehabilitation experience were surprising outcomes warranting further investigation. A further key observation in this study concerns the most common reason for rejection by both genders, "Prosthesis did not do what I need to do." This finding may be linked to barriers to treatment including fitting delays and receipt of quality training, as well as a comment on the current availability of appropriate solutions for the entire upper limb absence population.

This study sets the stage for further investigation as it relates to the continuity of care of individuals with upper limb absence. The importance of the quality and expertise of prosthetic and rehabilitation providers cannot be overstated, meanwhile, routine collection of objective and subjective outcomes is essential for establishing evidence-based care pathways and solution development. Furthermore evidence-based decision making enhances both the ability of individuals with upper limb difference to make informed decisions relating to their prosthetic experience and rehabilitation care and hence informs third party reimbursement policy.

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LONG-TERM FUNCTIONAL IMPROVEMENT WITH DEXTEROUS PROSTHETIC LIMB

Erin E. Sutton[1], Luke E. Osborn[1], Courtney W. Moran[1], Michelle J. Nordstrom[2], Paul F. Pasquina[2], Robert S. Armiger[1]

[1]Johns Hopkins University Applied Physics Laboratory, [2]Walter Reed National Military Medical Center

ABSTRACT

Advanced myoelectric prosthetic devices aim to restore functional capability after upper limb loss. However, studies of their functional impact have been mostly limited to short-term clinical studies which rely on assessments of simple manual tasks. Here we show that a longer term study can elucidate functional improvement and quantify how and when a prosthesis is used. A participant with transhumeral amputation and an osseo-integrated interface participated first in a ten-day study of functional capability with a highly prosthesis, the Modular Prosthetic Limb (MPL). A few months later, he took the MPL home and used it daily for 12 months. He returned to the laboratory for functional assessments every two months. We measured improved scores in Assessment of Capability with Myoelectric Control, Box and Blocks Test, and NASA Task Load Index over the course of the long-term phase. Only slight improvement was documented over the short-term clinic-based phase, which suggests that longer studies may be required to assess capability with highly dexterous prosthetic limbs. Additionally, the loads experienced by the limb in the home environment were much greater than during the laboratory visits, which suggests that the functional assessments do not capture the full spectrum of loads placed on a prosthesis during activities of daily living. Through the combination of functional outcome measures, on-board data logging, and long-term studies in the home environment, we are developing the capability to assess upper limb rehabilitation progress and device appropriateness.

INTRODUCTION

For people with upper limb loss, use of a prosthesis has been correlated with higher quality of life and rates of employment, but prosthesis abandonment persists. In recent studies, rejection rates range from 18% in the general US population [1] to 40% in the Veteran population [2]. Lack of function is the most widely reported cause of abandonment [1]–[3].

Myoelectric prostheses aim to restore functional capability, and commercially available terminal devices range from a powered hook to a multi-finger multi-grip prosthetic hand like the bebionic (Otto Bock, Berlin), capable of 14 selectable grips. The Modular Prosthetic Limb (MPL) is a research prototype with 17 independent actuators and infinitely configurable grips [4], [5]. Advanced devices like the MPL further extend the dexterous capability of upper limb prostheses. To match a given user's needs to a prosthesis in terms of dexterity, robustness, and usability, clinicians rely on functional outcome measures. However, these outcome measures have documented limitations [6], and even high quality measures neglect performance of domestic, everyday tasks. In contrast, long-term studies of prosthesis use in the home could elucidate how and when a prosthesis is used in activities of daily living and show functional progress. To date, this kind of study is rare and generally limited in duration to a few weeks or months [7]–[9]. Additionally, the time to train and master control of high degree of freedom prosthetic limbs is unknown.

We aimed to evaluate the functional capability of the MPL through a 12-month study. We collected continuous sensor data from over 100 sensors within the MPL including torque data from the device attachment site during daily use, and we intermittently assessed the user's functional progress with in-clinic outcome measures. Our results provide insight into the value of long-term evaluation of advanced upper limb prosthetic limbs.

METHODS

Our participant was a 63-year old male who underwent transhumeral amputation in 2007 secondary to cancer. The participant received targeted muscle reinnervation in 2012 and an osseo-integrated (OI) implant in 2015. Prior to starting the study, the participant had approximately 80 hours of experience with the MPL in a user-feedback and demonstration capacity. He provided informed consent, and all research activities were approved by the Institutional Review Board at Walter Reed National Military Medical Center.

We analysed the participant's data from two research efforts. First, he completed a ten day clinic-based study of the MPL with 11 other prosthesis users in May 2017. The study consisted of 12 laboratory training sessions of one to two hours each. Assessments of the MPL against his conventional prosthesis were conducted at the study's initiation, midpoint, and exit. The assessments were the Assessment of Capability with Myoelectric Control (ACMC) [10] and the Box and Blocks Test (BBT) [11]. The NASA Task Load Index (NASA-TLX) [12], a survey measure of mental load during a task, was performed after the ACMC and BBT. The second effort was the 12-month home study. During this phase, the participant was encouraged to wear the MPL for at least three hours a day during his activities of daily living. We evaluated his functional progress in clinical sessions every two months, and the same outcomes measures (ACMC, BBT, and NASA-TLX) were scored.

We also continuously monitored the loads on the OI interface throughout the home use phase of the study. Sensors mounted on the MPL measured the rotational torque (torsion) along the long axis of the humerus. Additional sensors measured the bending torque on the elbow joint about the elbow flexion/extension axis. We compared the loads on the arm in the clinical and home environments.

RESULTS

In the last session of the short-term, clinical phase of the study, the participant expressed that he had greater control of the MPL than at the start of the study, but that knew he could get better with more practice. This sentiment is reflected in the ACMC and BBT scores, which showed only slight improvement over the short term (Figure 1). The participant's prediction of his long-term improvement was correct. After approximately 100 days of home use, both the ACMC and BBT scores improved. Furthermore, the NASA-TLX results indicated that the mental load experienced by participant during the ACMC and BBT measures decreased over time.

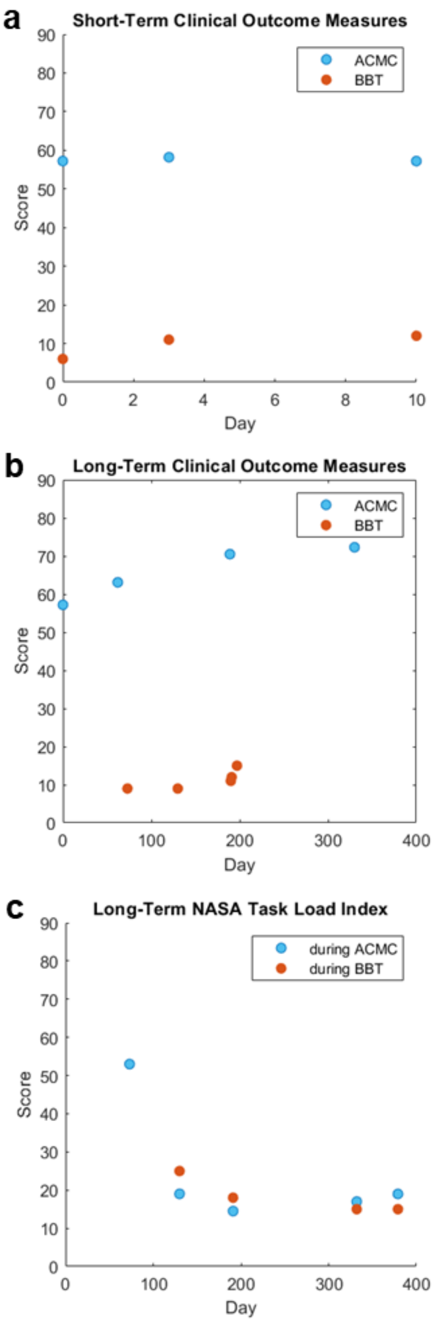

Figure 1. Clinical outcome measure scores. a) ACMC and BBT with the MPL during the short-term clinical study were relatively static. **b)** ACMC and BBT scores increased over the long-term, with the best scores recorded at the exit assessment. **c)** The NASA-TLX survey was administered after the ACMC and BBT. Lower scores on this measure indicate that lower mental load is required to complete a task.

Next, we compared the loads experienced during the clinical sessions of the 12-month home use study with the data recorded while the participant used the prosthesis at home. We recorded approximately 4.4 hours of wear a day during the take-home phase of the study and a total of 850 hours of data. We compared the elbow torque and OI torsion experienced by the limb during home use and during the six clinical sessions. Both elbow torque and OI torsion were much higher in the home environment than the clinical environment (Figure 2), which indicates that higher loads were placed on the MPL during the unstructured tasks of daily living than during the clinical functional assessments.

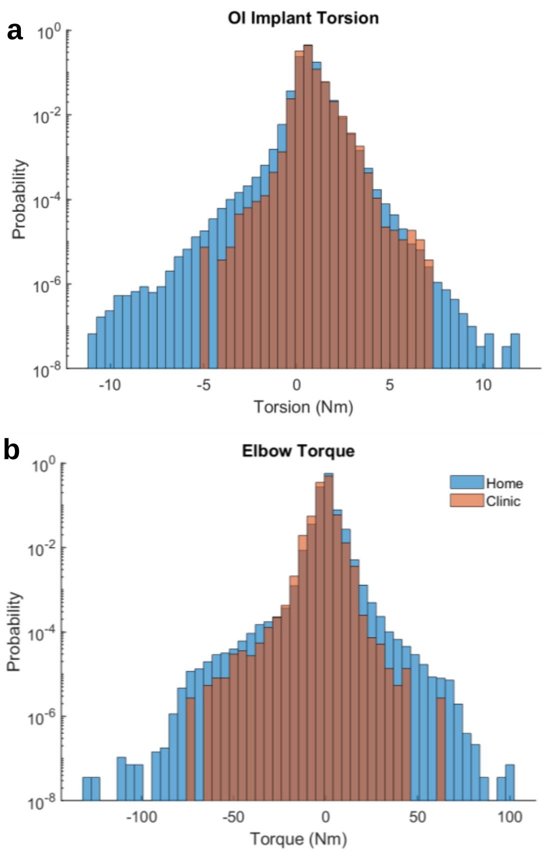

Figure 2. Histograms comparing loads on the MPL. For both graphs, the y-axis units are is the probability that a given data capture would be at a given torque. That is, higher probabilities correspond to more commonly measured torques. **a)** The torsion experienced at the OI interface was higher in the home environment (blue) than the clinical environment (orange). **b)** The elbow torque during all clinical sessions varied greatly from the torque recorded during home use of the MPL. The maximum torque magnitude in the clinical setting was 65 N-m compared to 135 N-m at home.

DISCUSSION

Our results suggest that a long-term study can capture functional progression with an advanced prosthesis even when progression is not evident over the short term. In the earlier study of 11 participants that compared the MPL with the participants' conventional prostheses, the MPL was found to out-perform the conventional prostheses, but gains in functional improvement over the two- to four-week study were unexpectedly low for some users. After that study, we expected that with increased wear time, a user's functional performance with the MPL would improve, and the longer term progress documented here supports this hypothesis. Additionally, the varied tasks demanded by the home environment could have contributed to increased capability over time. The participant reported frequent travel, daily meal preparation, and highly dexterous tasks like playing the piano. These tasks were demanding from a control perspective and likely contributed to the improvement in functional outcome scores. Furthermore, the increased functionality might have been a motivating factor to the participant's acceptance of the MPL, since his hours of continuous usage, the times he used the MPL without doffing, increased throughout the study.

The loads experienced by the limb during activities of daily living were much greater than during clinical assessments. This result has implications for the design requirements of new prostheses. Although we did not temporally map the higher torque loads to specific activities, the participant reported stressing activities like clearing his garden with power tools which could account for the higher torque. The high torsional values we recorded are consistent with data from intact limbs during advanced activities of daily living [13]. The frequent loads (50 N-m and 5 N-m for bending torque and torsion, respectively) are similar to loads previously reported from single session studies of OI transhumeral amputees [14]. Prosthesis users with OI implants have expressed concern about overloading the OI implant [14], and more data from active users like our participant could ease those concerns. Further studies could help shape the requirements for the safe use of an OI implant.

Through the combination of functional outcome measures, on-board data logging, and long-term studies in the home environment, we are developing the capability to assess upper limb rehabilitation progress and device appropriateness. In particular, our on-going work includes passive data collection methods such as wrist accelerometers and joint sensor data to monitor user performance.

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A MYOELECTRIC VIDEO GAME TRAINING PILOT STUDY: CHANGES IN CONTROL SIGNAL PROPERTIES

Carlos Martinez-Luna; Craig Kelly; Brianna Rozell;
Samuel Lambrecht; Andrew Horowitz; Todd Farrell, PhD

Liberating Technologies, Inc., A College Park Company

ABSTRACT

A myoelectric video game controller was developed which maps two-site upper-limb prosthesis control signals to mouse/keyboard commands via wireless Bluetooth. This Myo-Electric Gaming Interface (MEGI) is targeted for exercising and training clinically relevant control signal properties and strategies.

This study evaluated the effects of video game training on myoelectric control signal properties over a six-week period. A racing game was used to training proportional two-site myoelectric control using a differential control strategy and co-contractions. Signal amplitude maxima and distribution of control speeds were observed across the training of three pilot able-bodied subjects.

BACKGROUND

Myoelectric training is an important part of a new patient learning to control a prosthetic limb [1]. Upper limb prosthetic fittings are often not successful and eventually rejected because the prosthesis does not provide the functionality that the user expects [2]. Although this can be related to a number of different factors, it is often associated with the lack of adequate training. Myoelectric prosthetic users must develop control skills by exercising their remnant muscles. Without proper training, users often cannot reliably provide suitable myoelectric signals and can fatigue quickly, thus further negatively affecting their prosthetic performance. Many, if not most, upper limb amputees receive insufficient training on the use of their new prosthesis which may be due to healthcare funding, but also is likely due to the lack of appropriate training tools [2].

Dawson, et al. [1] provide a review of current commercial and research training technologies as well as their benefits and shortcomings. The technologies have been passable as the standard for training but have room for improvement, as they are too simplistic, not motivating, expensive, manufacturer specific, and cannot leave the clinic to allow for independent use by the amputee within their home. Requiring the device to be used in the clinic incurs substantial costs associated with the clinicians' time during the training process as well as the time and inconvenience for the patient to travel to and from the clinicians' office.

Figure 1: Myo-Electric Gaming Interface (MEGI) system.

The development of a new myoelectric training system strives to improve compliance of prosthetic upper-limb devices, and to build a more accessible and engaging approach to myoelectric signal training. Better patient outcomes will stem from prosthetic users being able to have more comprehensive and clinically relevant myoelectric training that is rewarding and entertaining.

The MEGI system is a myoelectric controller which leverages existing video game software, which are curated to elicit clinically relevant exercises [3].

MEGI promotes the use of proportional myoelectric control signals. In the prosthesis the amplitude of the signal controls speed, with larger signals creating faster movements. Good proportional control contributes to more efficiently controlling the prosthesis (especially for grasping and manipulation) and adjusting the grip strength of terminal devices [3]. MEGI trains proportional control by mapping myo-signal amplitude to video game controls like joysticks, where the myo-signal amplitude is mapped to how hard the joystick would be pressed in the game. In the case of racing games (Figures 1, 2) a lower signal is mapped to steering the vehicle gently and as the signal amplitude increases so does the degree of turning the steering wheel in the video game. Directional control is one of the most common control inputs in video games and, so mapping this directional control from the myo-signals encourages users to constantly work these muscles through a range of contraction intensities.

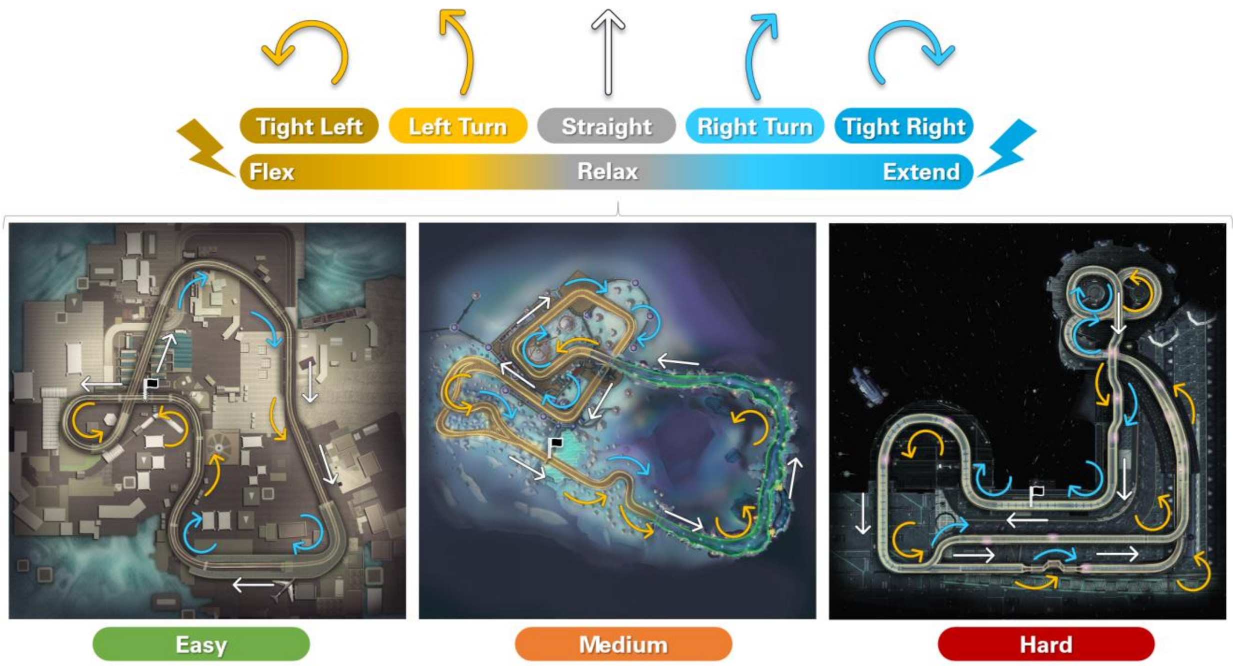

Figure 2: Representative training regimen racetrack courses, observed from a birds-eye view. TOP: Key of elicited differential myoelectric signal based on turn steering direction for right-handed wrist flexion/extension. MIDDLE: Representative courses containing balance of left/right turns. BOTTOM: Course difficulty with respect to road width, frequency of turns, and degree of turns.

Modern prostheses are made up of multiple devices; in upper-limb prosthetics one can have powered elbow, wrist, and hand prostheses. To control multiple devices with two-site myo, switching events are used, which require precise co-contraction or other pulse events. These signals are not typically used repeatedly, but when needed, they require precision and acute timing. Generation of co-contraction signals is one area that users of two-site myoelectric control often struggle with most. MEGI maps co-contraction events to binary button presses that are used periodically in game, but not repeatedly to the point of fatigue. In the racing game used in the study, co-contraction was mapped to using an in-game item for a speed boost or to fire a weapon.

Bimanual coordination is encouraged with the MEGI system mapping additional video game controls to a one-handed joystick peripheral used in the contralateral hand. In the racing game (Figure 1), this peripheral controls the gas pedal and brake.

METHODOLOGY

Subjects underwent informed consent and were instructed upon use of the MEGI device. Three able-bodied participants (2 male, 1 female, age 22±2), with limited previous experience with myo-control, completed the six-week study. Participants utilized two-site surface electromyography and differential control with wrist flexor and extensor muscles. All participants utilized dominant right arms for the MEGI system and their left hand for the joystick peripheral. Trials were conducted in a lab setting with investigators recording data.

Regimen

A commercial car racing game, *Sonic Racing*, for PC was used as the platform for training. The training regimen involved completing 3 racetrack courses per session and 3 sessions per week. Over the 6 weeks of testing, subjects completed a total of 54 courses. Subjects completed trainings every other weekday and rested on off days and weekends.

Eight different courses were selected which had a balanced level of left and right turns, corresponding to the eliciting of equal flexion and extension signals (Figure 2). Easy levels were used the first week of training, with more difficult levels introduced each week.

Data

The MEGI system recorded myoelectric signals with LTI "DC" electrodes acquired with a 10-bit analog-to-digital converter (ADC) onboard. The rectified myoelectric signal data was sampled at 30 Hz and included the differential signal (flexion minus extension) and deadband, as well as the raw signal of each channel. Lap times were recorded for each of 3 laps of each of the 3 courses in a training session.

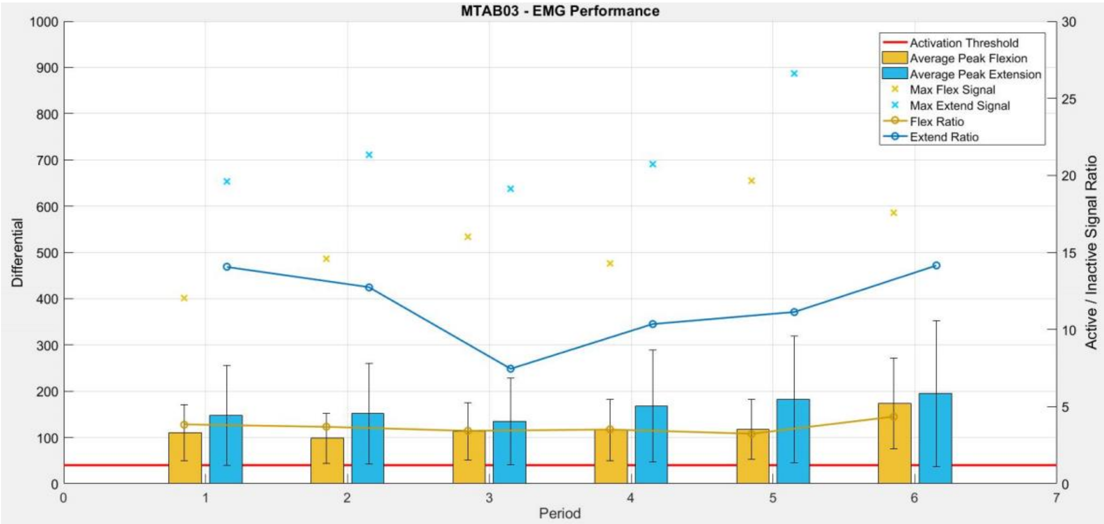

Figure 3: Myoelectric signal amplitude properties for representative subject (MTAB03).

RESULTS

The track and myo signal data collected during the training were analysed across the six weeks of training. Game performance was quantified across each track for the group. Signal characteristics such as maximum amplitude, peak flexion/extension, separation via flex/extend ratio during peak, and distribution of speeds were quantified for the dataset.

Previous studies have demonstrated that in video game training, participants can improve at the game itself while translation to improved prosthesis use can be less transferable [4,5]. This study focused on quantitative changes to the myo signals themselves and not prosthesis or functional testing.

Game Performance

In-game performance was quantified by measuring the lap time of every trial. Across the three subjects and eight courses, lap times decreased on average over the six test periods (Table 1).

Table 1: Average **v**ideo game racetrack completion times for all subjects (n = 3).

		Lap Time (sec)		Time/Period
Course		AVG	STDEV	Slope
Easy	A	57.7	1.2	-0.06
	B	55.8	2.0	-0.09
Medium	C	68.4	3.5	-0.14
	D	60.8	1.3	-0.04
	E	76.5	0.5	-0.08
Hard	F	78.2	2.0	-0.01
	G	92.4	3.2	-0.27
	H	94.2	3.5	-0.11

Amplitude

The flexion and extension signal amplitudes were tracked over time (Figure 3). Peak flexion and extension signals were calculated. As a measure of signal separation, the ratio of flexion to extension was calculated for each peak value. Flexion ratio was the peak value over the corresponding extension value, and vice versa. Generally, overall amplitude did not change outside of a 95% confidence interval.

Distribution

The range of myo signal amplitude elicited by a person can be correlated to the distribution of speeds that a prosthesis can be actuated. Similar research has evaluated prosthesis control performance by measuring the distribution of speeds across degrees of freedom in a prosthesis [6].

The distribution of myo signal amplitude was evaluated across the six test periods (Figure 4). For the differential signal, kurtosis (Eq. 1) was calculated to quantify the distribution at the tails, corresponding to the higher end of the flexion and extension ranges.

EQ 1. $$Kurtosis = \frac{1}{n}\sum_{i=1}^{n}\left(\frac{x_i - \bar{x}}{SD(x)}\right)^4$$

The hypothesis was that the myo-training would promote muscle growth and an increase in the range of the signals available. This could be quantified with a measure of kurtosis, indicating whether there is more signal distributed at the higher (flexion and extension) values, compared to the centre of the distribution. A lower kurtosis value correlates to a flatter distribution across the full range of amplitude.

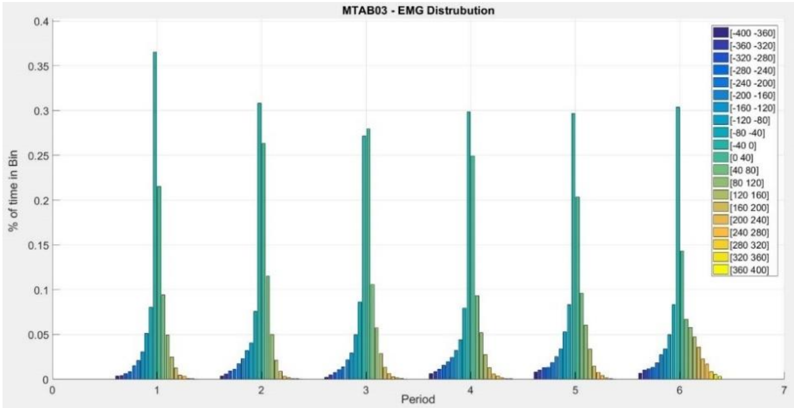

Figure 4: Distributions of myo signal amplitudes across the six weeks of training for one representative subject (MTAB03). Amplitude binned by analog-to-digital counts from a 10-bit DAC. Extension in blue, flexion in yellow.

Globally kurtosis values (Figure 5) are all greater than a value of 3, which indicate a leptokurtotic distribution which is defined as having fewer extreme values than a normal distribution. Across subjects, the trends in kurtosis over time seem inconsistent. Subject MTAB01 saw a nearly monotonic increase of kurtosis over time, while MTAB03 saw a decrease. Subject MTAB02 had a decrease in kurtosis after the first week, and then oscillated about a steady value for the remaining periods.

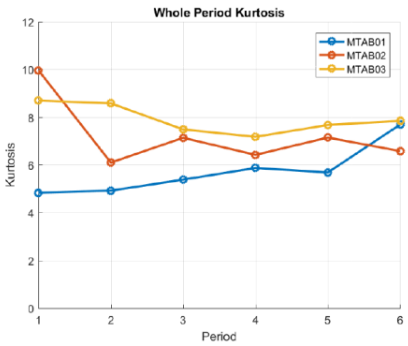

Figure 5: Kurtosis values over time for three subjects.

DISCUSSION

The signal amplitude and distribution data insofar are inconclusive. The distribution of electromyographic amplitude can be further evaluated in terms of uniformity and skewness. More signal distribution at higher amplitudes may not necessarily be indicative of good control. While a wide range of signal would correspond to more proportional control, some distribution at moderate values could be indicative of efficient control of the prosthesis. LTI is considering different measures of the myo-training separate from the game training itself. Standalone myo-signal tests and tracking tasks are under development, along with a suite of functional outcome tests.

Future Work

More work needs to be done to separate the learning effects and improvement of gameplay, and to quantify how that affects myoelectric control.

LTI is conducting the full study with research participants with upper-limb absence, over the course of multiple six-week periods, in order to evaluate changes in the target population and over longer terms of time.

ACKNOWLEDGEMENTS

This work is supported by the US Army Medical Research and Materiel Command under Contract Number W81XWH-18-C-0084. In the conduct of research where humans are the subjects, the investigators adhered to the policies regarding the protection of human subjects as prescribed by the Code of Federal Regulations (CFR) Title 45, Volume 1, Part46; Title 32, Chapter 1, part 219; and Title 21, Chapter 1, Part 50 (Protection of Human Subjects).

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PSYCHOMETRIC PERFOMANCE OF A 9-ITEM PROMIS UPPER EXTREMITY INSTRUMENT FOR PHYSICAL FUNCTION AMONG INDIVIDUALS WITH AMPUTATION

Dwiesha L. England[1], MS, Taavy A. Miller[1], MSPO, Phillip M. Stevens, Med[1,2], Shane R. Wurdeman, PhD[1,3]

[1]Department of Clinical and Scientific Affairs, Hanger Clinic, Austin, TX, USA,
[2]School of Medicine, University of Utah, Salt Lake City, UT, USA
[3]Department of Biomechanics, University of Nebraska, Omaha, NE, USA

ABSTRACT

The use of outcome measures are often a policy-driven requirement when assessing the efficacy of clinical care in several patient populations including prosthesis users. A recent review of upper extremity instruments described the PROMIS measures as a “potential improvement to current practice”[1]. Thus, the PROMIS-9 UE was developed from the PROMIS UE item bank to assess psychometric performance among individuals with UE amputation. Performance testing was achieved by evaluating structural and known-groups validity, reliability and differential item functioning (DIF) among participants. To be structurally valid, the assumptions of unidimensionality (one dominant factor obtained), local independence (i.e. all LD $\chi 2<10$), monotonicity (scalable coefficient for the full scale equates to 0.57) and good model fit (p-values>0.006 for all items) were confirmed. The graded response model results, for the item difficulty parameter, revealed that the nine items were covering low to moderate levels of physical function. Known-groups analysis demonstrated that prosthesis users had significantly higher levels of physical function compared to non-user (p=0.039). Lastly, the PROMIS-9 UE had adequate item response theory (IRT) reliability, 0.9, and no age DIF were found. Although there is a need for more challenging questions, the PROMIS-9 UE psychometrically performed well supporting its continued utilization for individuals with low to moderate levels of physical functioning.

INTRODUCTION

Prominently utilized upper extremity (UE) physical function instruments, that predate the establishment of the PROMIS physical function UE item bank, report having limitations such as ceiling effects, non-unidimensional factor structure, or lengthiness [2], [3]. To overcome these limitations, the PROMIS group developed fixed length short forms and computer adaptive test from validated item banks across several domains including pain, anxiety and physical function. The PROMIS v2.0 UE physical function item bank allows [4] content experts to create a customized short form by selecting items clinically relevant to their targeted population and subsequently test its performance in a clinical setting. A recent study found that the PROMIS UE item bank had good psychometric properties such as adequate structural validity, sufficient differential item function and good reliability among individuals with upper limb complaints [5]. To build on this body of evidence, it was hypothesized that a customized 9-item measure, PROMIS-9 UE, chosen from the PROMIS v2.0 UE item bank, will also perform well within a specified population of individuals with upper extremity amputation.

METHOD

Study design

Patients with UE amputation across the United States completed the PROMIS-9 UE measure and demographic data during a routine visit with their prosthetist. Retrospective chart review of cross-sectional data was used to determine the psychometric performance of the PROMIS-9 UE.

Subjects

A database containing 269 patients were reviewed. To be included in the analysis, individuals had to be 18 years and older, have received an upper extremity amputation and have completed a PROMIS-9 UE questionnaire.

Statistical Analysis

All statistical analyses were performed using IRTPRO (version 4.1) and R (version 3.6.1). Patients’ demographic data were described using sample means, standard deviations and percent proportions.

The structural validity of the PROMIS-9 UE was assessed by evaluating three IRT assumptions before fitting a graded response model. These assumptions include: 1) unidimensionality, 2) local independence and 3) monotonicity [6]. Unidimensionality is define as the instrument ability to measure one domain, for this current study, physical function. Exploratory factor analysis was used to determine if the PROMIS-9 UE had one factor or a dominant first factor. Local independence dictates that there should be no association between items, after controlling for the measured trait. This was verified using IRTPRO's local dependence chi-square statistics (LD $\chi 2$). If any LD $\chi 2$ value exceeded 10 then local independence was violated [7]. Lastly, monotonicity occurred when the probability of selecting a higher response category increases with the levels of the measured trait. The R-package Mokken (version 2.8.11) was used to verify whether monotonicity was held for the PROMIS-9 UE instrument. If all three assumptions were met, results obtained from the logistic graded response model with S-$\chi 2$ can be interpreted. From the model, p-values less than 0.006 are suggestive of poor model fit and a wide range for the item difficulty parameter is suggestive of good coverage.

Known-groups Analysis

Known-groups analysis was used to assess differences in physical functioning T-scores for prosthesis users versus non-prosthesis users. This was carried out using an independent samples t-test. T-scores were obtained from HealthMeasures.net scoring service.

DIF and Reliability

When the influence of age, gender or education status impacts an individual's response to an item category, then DIF has occurred. Items flagged for DIF can add noise to the instrument and some studies recommended that non-relevant items with significant DIF be excluded. DIF was assess using IRTPRO (version 4.1). Reliability evaluates the instrument's capabilities to precisely measure the domain of physical function. Traditional Cronbach's alpha gives the reliability for the entire instrument while the IRTPRO reliability gives the precision for individual values of T-score within the scale.

RESULTS

After removing patients with incomplete PROMIS-9 UE data, a convenience sample of 239 individuals was retained in the final analysis. Over 70% of the population were male, 45% were transradial and 63 % were prosthesis users at the time of the survey (table 1).

Table 1: Patients' Characteristics

	Count (n)	%
Total Sample	239	100
Gender, male	170	71
Education, college degree	143	60
Employed, yes	106	44
Acquired amputation, yes	171	72
Amputation Level		
Transhumeral/elbow	40	17
Transradial/wrist	107	45
Prosthesis user, yes	150	63
	Mean	**SD**
Age of participants (yrs)	48	16
Use of prosthesis (hrs/day)	9	5.3
PROMIS-9 UE T-scores	29.6	9.8

Structural validity

Unidimensionality analysis revealed that physical function was a dominant factor for the PROMIS-9 UE. None of the items violated the assumption of local independence as all LD$\chi 2$ values had a magnitude less than 10. Also, the assumption of monotonicity was met because the scalability coefficient for the full scale (0.565) exceeded the minimum value of 0.5. Model results indicated that none of the items were poorly fitted (p >0.006). The item difficulty level of the scale ranged from -1.44 to 1.34 suggesting low to moderate coverage for physical function

Known Group Validity

As expected, prosthesis users had significantly higher t-scores than non-prosthesis users (p=0.039).

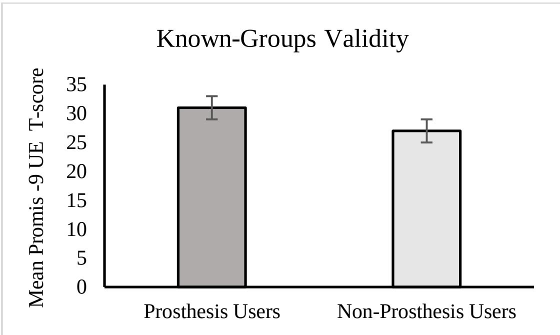

Figure 1: Prosthesis users had significantly higher physical functioning scores when compared to non-users.

Reliability

The average IRT reliability estimate for T-scores values found for the middle (28-70) of the scale was 0.9 indicating adequate reliability. Figure 2 showed that as the information increased the reliability simultaneously increased. Similarly, the traditional Cronbach's alpha analysis revealed adequate reliability value of 0.93 for the entire scale.

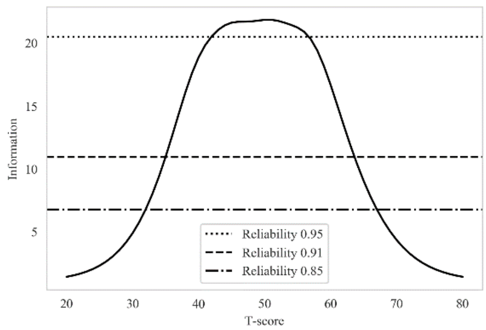

Figure 2: Information plotted across range of T-scores. The T-score range of 29-70 has the greatest level of precision. The reliability reference line of 0.95 correspond to an information magnitude of 10.

DISCUSSION

The purpose of this study was to examine the psychometric performance of the PROMIS-9 UE among individuals with UE amputation and this was achieved. Study results demonstrated no significant violation of validity, reliability and differential item functioning.

Hung et al. concluded that the PROMIS v1.2 UE item bank for physical function was structurally valid for individuals among upper limb complaint and further noted that more challenging questions are needed to capture higher functioning individuals. Similarly, our graded response model reported strong performance among UE amputees and also reaffirm the need for the addition of more difficult questions to the existing item bank. For example, if more challenging questions are added to the bank, then the two of the four items in the PROMIS-9 UE with similar range of item difficulty could be replaced with more challenging ones. Yet, the need for refinement does not preclude the administration of this instrument at baseline assessment and perhaps follow up visits for patients' with low to moderate levels of functionality.

This study is not without limitation. Future study should consider the performance of the PROMIS-9 UE with longitudinal data. This will demonstrate how well the instrument can track changes in patients' functional status. Lastly, future study should consider the impact of device type on the increase or decrease of patients' physical functioning T-Scores.

In conclusion, although challenging questions are needed to provide coverage for individuals with high actively levels, the PROMIS-9 UE is psychometrically sound and can be administer to patients with low to moderate physical function activity level.

ACKNOWLEDGEMENTS

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SURVEY OF BILATERAL UPPER LIMB PROSTHETIC USERS

Harold H. Sears, PhD; Kim Doolan BS; Denise Keenan OTR, CHT

Unaffiliated

ABSTRACT

Bilateral upper limb loss (BiULL) is perhaps the greatest challenge for upper limb prosthetic care, now more than ever, as we witness the increase of sepsis as a major cause of multiple limb loss. This small-n survey has recruited 28 individuals with BiULL, 27 of whom are prosthesis wearers. 12 of the 28 lost four limbs to sepsis; 17 of the 27 prosthesis wearers use body-powered hooks, six use electric hooks, and four use electric hands as their dominant terminal device. Secondary prosthetic use is also included, when the secondary prosthetic set was used for 10% or more of total activities.

The survey used person-to-person interviews to compile detailed data about how tasks are performed, how many tasks are performed, etc. A detailed picture is painted from this data, including the functionality and independence achieved by many in this population, and the needs expressed for improvements in their devices of choice, and the care they receive. For example, the indications for improvements needed emphasized greater dependability, and greater grip security. Ratings of prosthetic features illuminated shortcomings in training especially.

The information should be useful for clinical guidance, but also to help guide the development of future prosthetic devices, as well as set an example for how a small-scale study can collect useful data about the use of prosthetic devices, without a large grant or large institutional sponsorship.

A. BACKGROUND

The bilateral upper limb loss (BiULL) individual presents perhaps the greatest challenge in UL rehabilitation. Since there is a dearth of information in the literature about the actual needs of this small but important population, this small study hopes to contribute relevant knowledge towards both the clinical and development needs that exist. It is also expected that wearers with BiULL use their prostheses in the same ways as wearers with unilateral limb loss (LL), i.e., what is needed by the small group in this study is also going to be needed by the larger population with unilateral LL.

From previous experience with similar surveys[1,2] the in-depth information available from personal interviews with prosthesis users has been used successfully to focus on prosthetic needs. A large segment of the entire population of BiULL individuals may be nearly impossible to recruit, but gathering in-depth information from the 28 subjects in this small study provides a wealth of information (about the details of prosthetic use) that would be more difficult with a large-n study.

Methods: The data collected in this survey seeks to document all the ways that BiULL persons use their variety of prosthetic devices, and the ways they are still limited by those devices. Direct interviews with all subjects, either in person or by telephone, allows the open-ended discussion necessary to collect the breadth of information sought.

The simple assumptions, upon which the study is based include:

- No research grant, thus no delays for proposal writing and funding.
- No oversight by a large institution, thus less staff to coordinate, less “red tape”, etc.
- The authors each have 30+ years’ experience in the prosthetic field, working as therapist, prosthetic coordinator, and engineer/manager. The first author has conducted earlier surveys with published results.
- Data is collected directly from subjects within the population with BiULL, who are directly recruited.
- This project hopefully can set an example others could follow. The highest priority is to gather data from consumers – a priority recognized by the limb loss community as well, in the 2018 Amputee Coalition study[3] which cited the great need for outcomes reflecting the actual needs and priorities of the limb loss population.

Recruitment: Many (approximately half of the 28 subjects) were recruited at the Fifth Skills for Life (SFL5) Workshop, attended by over 70 persons with BiULL, held in Houston, TX, in October 2018.

Institutional Review Board (IRB): The protocols and informed consent form were reviewed by a certified private IRB (Ethical and Independent Review Services, Corte Madera, CA), and the study was judged to be exempt from IRB oversight, citing no risk to subjects. Subjects were not compensated.

RESULTS:

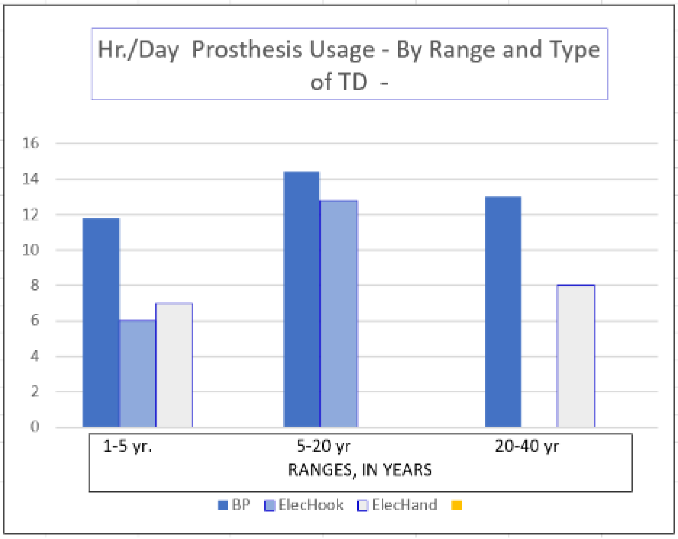

Figure 1 – Average daily usage reported by the subjects, in the three nearly equal ranges. Again, BP usage is on average very high, and only approached by Electric Hooks in the middle range.

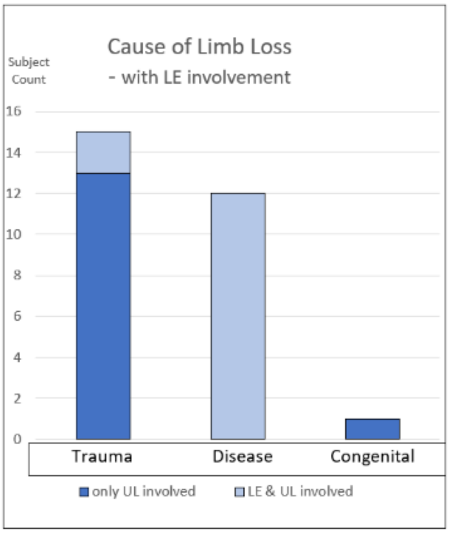

Figure 2 – Cause of limb loss, showing the significance of disease-caused limb loss (sepsis, in all cases, also causing LE loss).[4]

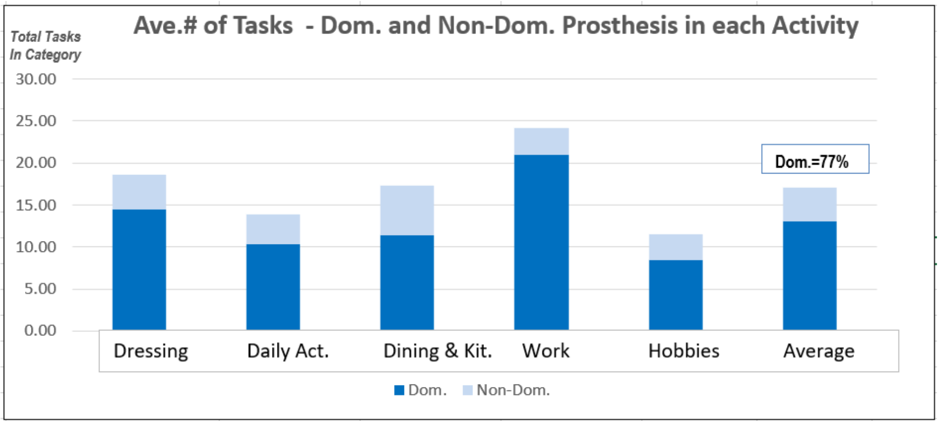

Figure 3 – Summing the total tasks performed in each of five categories, shows the dominant side consistently is the most heavily used - 77% on average . Data includes all TDs, all loss levels. On average 85 different tasks are performed, some many times each day, so total tasks are underestimated.

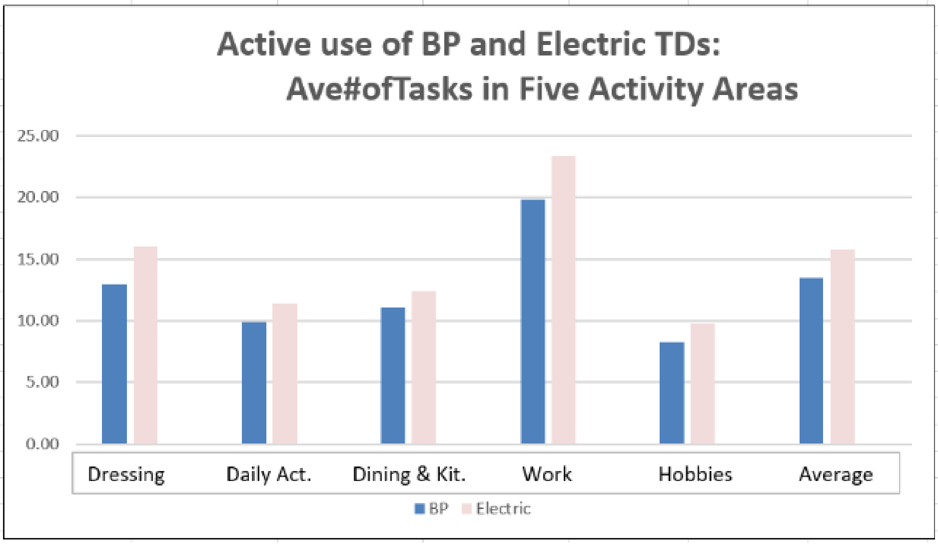

Figure 4 – The total number of tasks tallied in each of the categories, including the average of all five. In this case the electric TDs tasks (both hook and hand) are slightly higher, but the difference is not statistically significant.

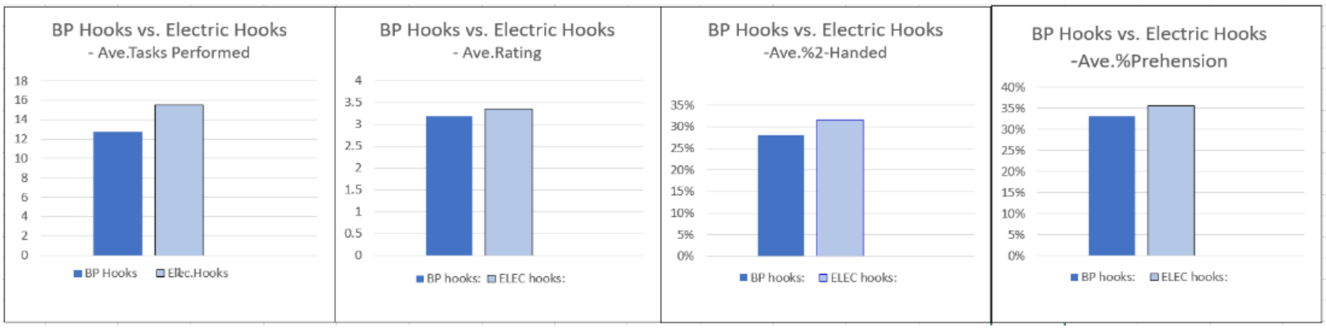

Figure 5 – Comparison of BP Hooks (n=16) vs. Electric Hooks(n=9). Data is average of five activities (for total tasks multiply by five) and includes both primary and secondary prostheses, if used. Charts are i to iv, left to right:

i. *Tasks reported- Total tasks (repetitions of tasks not included) (ElecHooks +22% higher).*
ii. *Average rating, 4=A, 3=B, 2=C, 1=D, 0=F (ElecHooks +5% higher).*
iii. *Average percent of 2-handed tasks (ElecHooks +13% higher).*
iv. *Average percent of tasks using prehension (ElecHooks +8% higher).*

Other survey results included the subjects' ratings of prostheses in specific features, which can help to explain some of the results presented in Figure 4 and 5, e.g., electric hooks were rated higher in grip security, which was a very high priority for all the surveyed group. "Improvements Desired" was solicited from subjects, and produced a high amount of data, listing 15 specific shortcomings of present devices, mostly centered on the terminal devices. The clear areas of most need could be generalized as: *Durability* (four distinct areas were cited), and *Grip Security* (including hand and hooks, electric and BP). "Impact of Training" was also graded. Electric prostheses graded their training a D+; BP prostheses graded training a C. In addition to the prosthetic devices used by subjects, 'Other Assistive Devices' (in 10 categories), were very important to nearly all subjects, and are used in many diverse activities, including: household activities, driving, bathing, eating, computer/phone functions, and sports.

Conclusions from the data

1. Functional capabilities of the surveyed group are on average very high – and notably, for all the prosthetic choices: e.g., BP Hooks, Electric Hooks, and Electric Hands. The majority have chosen body-powered hook TDs, but for subjects whose experience is within the last 20 years, the group is nearly equally divided between electric and body-powered devices.

2. The functional needs expressed, considering all devices, are led by *better dependability* and *better grip security.* Other needs included better range of motion, water resistance, comfort, and lower weight. Generalizing, the surveyed group appreciates what they have accomplished, but they know improvements *could give them better function*- as long as the *dependability, versatility, and affordability* they value are not sacrificed. Choosing the right device for the individual need not be haphazard. Careful evaluation and trial fitting could give patients and caregivers better choices. [5,6]
3. Prosthetic use by this group shows: very high use of the *dominant side* prosthesis over the non-dominant side (75% vs. 25%), as well as very high use of *passive function,* over prehension functions (65% vs. 35%).
4. Other contributions to function:
   a. Additional assistive devices, of a wide variety from a home-made zipper holder to driving rings, and clothes pins (13 different categories are enumerated).
   b. Consumer electronics (phones, tablets, computers, etc.) and Automotive electronics aid this group immensely.

Indications for additional study about the BiULL population.

1. The priority for improvements in *dependability* and *grip security* were high in this survey of 28 persons with BiLL.
   Larger studies (or focused small studies) could *verify* these conclusions, and could also be more *specific in comparing types of hooks and hands*, control options, or the impacts of important variables such as *expert prosthetic care* and the *center-of-excellence* approach*, early fitting and training, mental health* services and other technologies.
2. *Training* clearly is an area of great potential- but exactly how to improve training must be studied seriously. A few possibilities include (but are not limited to):
   a. *Telehealth* shows potential for leveraging the impact of expert therapists to provide wider access to skilled therapy, custom training for clients, and training for therapists in specific skills.[7]
   b. Internet links such as You Tube video of skilled users, are widely accessed consumers, and could supplement training for therapists also.
3. Focused evaluation studies of specific prosthetic TDs would help consumers to understand the pros and cons of new (or old) devices, before making expensive choices. *Cost-benefit analysis* is difficult in prosthetics, but could be developed as a benefit to consumers, and prescribers as well.

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TECHNOLOGY TO MONITOR EVERYDAY UPPER-LIMB PROSTHESIS USE – A REVIEW

Alix Chadwell[1], Dafne Zuleima Morgado-Ramirez[2], Laurence Kenney[1], Laura Diment[3], Encarna Mico-Amigo[3], Malcolm Granat[1], Alex Dickinson[3]

[1]University of Salford, UK, [2]University College London, UK, [3]University of Southampton, UK

ABSTRACT

Real-world monitoring offers an objective way of exploring the everyday wear and use of upper-limb prostheses. To inform future developments in this field, a systematic literature review was undertaken, highlighting studies that monitored the activity of prosthesis-users during daily-living. Nine papers relating to the upper-limb were identified, and sixty relating to the lower-limb. Here we concentrate on the ways in which technologies have been utilised to assess the use of upper-limb prosthesis, whilst also drawing on the findings of the broader review to highlight potential uses of these measures, alongside the benefits and disadvantages of different approaches.

INTRODUCTION

If the benefits associated with wearing a prosthesis are outweighed by the drawbacks, then a person may choose not to wear or use it [1,2]. Additional complexity, weight and cost associated with prosthetic prehensile function is only of sufficient value if it is used in everyday life. Clearly, these issues around wear and use are context (e.g. time/setting) specific and may vary person to person. However, until recently, the primary methods of determining how upper-limb prostheses were worn and used on a day-to-day basis was through self-report and examination of the prosthesis (e.g. a worn-out cosmetic glove or mechanism). Over the past 5-6 years, researchers have begun to use technology (e.g. sensors on, or in the prosthesis) to objectively assess upper-limb prosthesis wear and use once the person leaves the clinic.

Although monitoring of real-world wear and use is a relatively new approach to upper-limb assessment, the first papers reporting activity monitoring in people with lower-limb absence were published in the 1990's. By understanding how researchers have used real world monitoring to assess lower-limb prosthesis users, as well as the relative merits of the different approaches, it may be possible to guide the development of appropriate approaches to the evaluation of upper-limb outcome measures.

Here we present the results of a literature review which explored the ways in which technology has been used to monitor everyday prosthesis use. The findings of studies using real world monitoring techniques in upper limb applications will be presented, together with potential lessons to be learnt from the lower-limb field. Finally, conclusions will be drawn as to future work.

METHODOLOGY

Five databases (MedLine, Web of Science, Scopus, CINAHL and EMBASE) were systematically searched to identify all papers published up to 1st November 2019. The search employed three groups of keywords as detailed in Figure 1.

Real-world activity:

"daily living" OR "free living" OR "daily life" OR "real world" OR activit* OR mobility OR "prosthetic use" OR "home use" OR "real life" OR "daily use"

AND

Population of interest:

"artificial limb" OR "artificial leg" OR "artificial arm" OR (prosthe* OR amput* AND (limb OR leg OR arm OR hand OR wrist OR elbow OR foot OR ankle OR knee OR transradial OR trans-radial OR transhumeral OR trans-humeral OR transtibial OR trans-tibial OR transfemoral OR trans-femoral))

AND

Sensor for monitoring activity:

actimetry OR sensor OR monitor* OR "inertial measurement unit" OR IMU OR acceleromet* OR gyroscope OR magnetometer OR "global positioning system" OR GPS OR "step count" OR pedometer OR "cadence" OR "steps/" OR "steps per"

Figure 1: Search terms employed to identify all studies that monitored the activity of prosthesis-users during daily-living.

Only papers which reported first-hand on sensor-based monitoring of people with prostheses in a community setting (i.e. outside the lab or clinic) were included in the final review. For all included papers, reference lists and forward citation reports from each database were consulted in order to identify additional relevant articles that were not found in the automatic search.

RESULTS

The search returned 2793 papers across the 5 databases. After removing duplicates, 1716 were screened by title and abstract; of these, five papers relating to the upper-limb were identified as relevant [3-7]. Analysis of references and citations highlighted four additional upper-limb papers [8-11] (Total = nine papers). For comparison, 60 papers relating to the lower-limb were identified.

With respect to monitoring upper-limb use, four research areas were identified:

(1) Use of wrist-worn accelerometers to measure aspects of symmetry in upper limb activity and prosthesis wear time [3-6]

(2) Use of head-mounted video cameras to generate grasp taxonomies [7,8]

(3) Use of on-board sensing to evaluate choice of grasp [9]

(4) Use of on-board sensing to evaluate the use of a sensory feedback system and the number of grasp events [10,11]

It is worth noting that during the review process, five other studies were identified, however these were excluded from the main review because they assessed upper-limb activity without clarifying whether a prosthesis was worn at the time [12,13], or because they were only undertaken as lab-based studies [14-16]. Any future community-based applications of these methods would be of interest.

When considering all 69 papers (upper- and lower-limb), there has been a large amount of growth in publications over the past 10 years (Figure 2). Most studies recorded data for between one and two weeks (Table 1). Studies lasting for less than a week were generally those concentrating on the development of devices and algorithms, whilst studies lasting for more than one month were mostly intervention-based. Studies that compared activity monitoring to clinical scores or that compared populations typically used a 7-day protocol. Only three studies lasted for longer than three months.

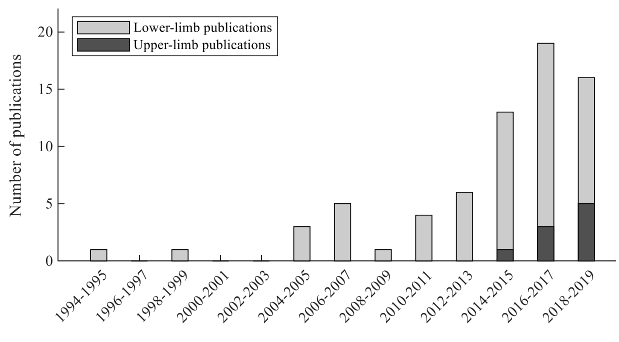

Figure 2: Number of publications per year (grouped into 2-year bins). 9 upper- and 60 lower-limb publications were published during this period.

Table 1: Recording period for studies split by the main focus of the manuscript.

Recording period	**Number of studies in each cateory**			
	Algorithms	***Clinical Scores***	***Interventions***	***Populations***
<7 days	6	1	1	1
7-14 days	5	13	15	6
15-30 days	1	2	5	1
31-90 days	0	2	5	1
>90 days	0	2	1	0

DISCUSSION

Although only 9 studies addressed the everyday assessment of upper-limb activity using activity monitoring methods, within the lower-limb field, these methods were observed to have increased in popularity over the past 10 years. Results suggest that upper-limb monitoring within prosthetics is approximately 10 years behind the lower-limb field, and as such we anticipate an increase in the use of real-world monitoring in the coming years.

A substantial proportion of the lower-limb studies focused on comparing prosthetic components such as different designs of foot spring. By introducing activity monitoring techniques into the upper-limb field, it will be possible to objectively compare how different types of prosthetic hand design, control methods, or socket designs impact on everyday wear and use. Other key uses of these methods in the lower-limb field included lifestyle interventions and to allow comparisons between populations. Additionally, several studies looked at comparing activity level against various clinical scores (for example K-levels). It would be interesting to use real world monitoring techniques in the upper-limb to evaluate the effects of user training methods.

The upper-limb papers identified in this review reported data on either the movements of the arm(s) (using accelerometers), or the number/types of grasps used in daily life (using video cameras or on-board processors). Neither of these measures on their own provide a complete understanding of both when the prosthesis is worn and how much it is used. For a person with an upper-limb prosthesis there are many aspects of use to consider, including: Is the arm used? Are the arm movements similar to those of an anatomical arm or do they reflect compensatory movements? Are the active capabilities of the hand, such as grasping, being used and if so, to what extent? Although the field is in its infancy, many of these issues are beginning to be explored by different groups and hence there is great potential to combine techniques. For example, by combining accelerometry for the detection of arm movements with recordings of grip choice and frequency of use, comparisons could be made with studies of upper limb activity in anatomically intact populations, between users of different types of prosthesis, or with people with different upper-limb impairments. Further by comparing measures such as 'system on-time' against prosthesis wear time it is possible to understand the value of advanced systems such as sensory feedback [9,10].

Prosthesis wear time is a key outcome with respect to the upper-limb, as if the user does not find the prosthesis to be of sufficient value, then it will not be worn. Consequently, reporting of prosthesis wear time is much more common in these studies than those relating to the lower-limb, where non-wear may be less of a choice with movement requiring crutches or a wheelchair when the prosthesis is not worn, thus greatly reducing functionality. Although algorithms for the automatic detection of upper limb prosthesis wear/non-wear have been developed [5,6], further validation is needed before these can be widely accepted.

This review suggests we are still some way off properly understanding real world behaviours of prosthesis users and the factors which influence them, however, many opportunities for development have also been highlighted. With growing numbers of low-cost 3D printed prosthetic hands becoming available, and the high cost of some advanced technologies, these objective methods of assessment offer the potential for significantly improving our understanding of the value, or otherwise of prostheses to users. As with all 'real world' monitoring technologies, ethical issues will also need to be addressed and there are several interesting discussions on these issues, which become more complex with increasing invasiveness of prosthetic technologies [17]. Such approaches would be helped by the development of agreed standards on which data should be recorded and how these should be represented, which in turn may assist with evidence-based commissioning and prescription of upper-limb prostheses.

ACKNOWLEDGEMENTS

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UNDERSTANDING HOW HARNESSING AFFECTS A USER'S WORKSPACE

Alix Chadwell[1], Laurence Kenney[1], Dave Howard[1], Robert Ssekitoleko[2], Brenda Nakandi[2], John Head[1]
[1]*University of Salford, UK,* [2]*Makerere University, Uganda*

ABSTRACT

Despite the fundamental importance of reachable workspace in upper-limb prosthetics, to date there have been no studies on this aspect. We have developed a methodology to quantify the reduction in the reachable volume of body-powered prosthesis users due to harness setup, and to record the range-of-motion of the prehensor at a series of locations within the workspace. For this proof-of-concept study ten anatomically intact participants were assessed using a prosthesis simulator. Data was collected using a 3D motion capture system and an electronic goniometer. The harness/cable reduced the reachable workspace by 15-62% with participants struggling to reach across the body and above the head. Across all arm postures assessed, participants were only able to achieve full prehensor range-of-motion in 9%. The methodologies could be useful in guiding the setup of body powered prostheses and in the evaluation of future designs of both body-powered and myoelectric prostheses.

INTRODUCTION

Reachable workspace is a key measure within the fields of upper-limb rehabilitation [1], [2] and robotics [3], with reduced workspace being shown to have a negative correlation with quality of life [4]. For upper-limb prosthesis users, the reachable workspace may be reduced due to a reduction in the degrees of freedom available in each of the joints (e.g. a prosthetic socket restricting full flexion of the elbow).

For a user of a body-powered prosthesis, the cable routing of the control harness can cause further restrictions, sometimes preventing the user from reaching certain parts of the workspace. Additionally, the ability of the user to fully exploit the 'Mechanical aperture RoM' of the prehensor may also be affected by the harness setup. Increasing the length of the cable during setup to increase the size of the reachable workspace, could negatively impact on the achievable aperture Range of Motion (RoM) so that the user cannot fully close a voluntary closing (VC) terminal device in some arm postures. Conversely, decreasing the length during setup, to ensure full closure is always possible, could prevent the user from fully opening the device in some arm postures, and from reaching certain parts of the workspace.

To reflect the need to find a compromise, various clinical guidelines have been developed; however, these vary and are somewhat vaguely worded. Further, whether any of the resulting setups are optimal in any formal sense is not known. Many prosthetists will rely on their own experience when setting up the harness system, and it is not known what the most common setups are.

The extent of the workspace limitations and the implications on function have not been explored. Until we have methods to quantify these limitations, design and setup decisions will be difficult to justify. Therefore, the aims of this proof-of-concept study were to develop suitable methods with which to quantify the limitations on both reachable workspace and the ability to fully exploit the 'Mechanical aperture RoM' of the prehensor within this space.

METHODOLOGY

Ten healthy anatomically intact adults were recruited. Ethical approval for the study was granted by the University of Salford Health Research Ethics committee (REF: HSR1819-050) and informed consent was gained from all participants. Participants were assessed using a TRS body-powered prosthesis simulator, consisting of a right-handed wrist brace, a figure-of-9 (P-loop) harness, and a TRS Voluntary Closing GRIP3 prehensor. Motion data from body-worn and prosthesis-mounted reflective markers were captured at 100Hz using a 13 Oqus camera system (Qualisys, Gothenburg, Sweden), and an electronic goniometer (SG75, Biometrics Ltd) was attached across the mobile 'thumb' of the prehensor to measure prehensor aperture (opening and closing).

To assess the impact of the harness on the reachable workspace, participants attempted a series of arm sweeps around the body under two conditions: unharnessed and harnessed. To capture the reachable workspace, participants were asked to sweep their hand through 9 arcs with the elbow fully extended (note that the contralateral (left) shoulder remained in a neutral position throughout). These arcs were parallel to the frontal, sagittal, and transverse planes as shown in Figure 1.

The next part of the experiment was to evaluate the extent to which the participant could open and close the prehensor within their reachable workspace. Whilst holding the prehensor in a range of pre-specified locations around the body, participants were asked to open and close it as far as possible by only abducting and adducting the contralateral (left) shoulder.

3D marker co-ordinates from the prosthetic 'finger', right shoulder, and a cluster of three markers on the sternum were exported from Qualisys, and data processing and analysis was undertaken using Matlab (Mathworks Ltd). This included:

- Filtering of 3D co-ordinates and goniometer data.
- Rotation of 3D co-ordinates from the lab frame into a co-ordinate frame based on the sternum.
- *Reachable Workspace:* Calculation of the convex hull surrounding the 3D co-ordinates from the 'finger' marker and the sternum origin marker using the Matlab alpha shape function.
- *Reachable Workspace:* Removal of the surface of the convex hull behind the person's back which joined the extremes of the arc sweeps. This was replaced by a surface joining these perimeter points to the sternum (Figure 2). The removed space corresponds to an area of the volume which the participant was unable to reach, thus overestimating the reachable volume.
- *Reachable Workspace:* Volume calculated.
- *Control Over Prehensor Aperture:* Grouping of hand positions into segments around the body.
- *Control Over Prehensor Aperture:* Mean Achievable aperture Range of Motion calculated for each segment.
- *Control Over Prehensor Aperture:* Results presented according to 8 segments around the body, 5 segments up/down the body, and 3 segments radially away from the body. All segments centered on the right shoulder.

RESULTS

Across all ten subjects, the harnessed reachable volume was approximately 70% of the unharnessed volume. At best there was a 15% reduction in the reachable volume when wearing the harness, and at worst a 62% reduction. Figure 3 shows example data from a participant with a large reduction in their reachable workspace (unharnessed volume = 1.25 m^3, harnessed volume = 0.49 m^3) as viewed from the front. When the control harness was connected, this participant struggled to reach their arm above the horizontal and across the body to the left-hand side.

All participants found it harder to open the prehensor in postures where the arm was crossed over to the left side of the body or when the arm was higher than the sternum as the harness was too tight to achieve full opening. Some participants also struggled to close the prehensor when the arm was on the right-hand side of the body as the harness became too slack. For most participants, as the arm moved down the body, the achievable aperture RoM increased. However, for some the increased slack in the system meant that the cable length increased to a level where they struggled to close the prehensor in the lower segments. When participants operated the prehensor near to their chest, very few were able to open the prehensor beyond 50% aperture. As they extended their arm away from the body, the achievable aperture RoM generally increased.

Participants were only able to achieve the full 'Mechanical aperture RoM' in 9% of postures assessed. In 38% the achievable aperture RoM was <=50% of the 'Mechanical aperture RoM'; in ~2/3 of these the participant struggled to open the prehensor and in the other ~1/3 they struggled to close the prehensor.

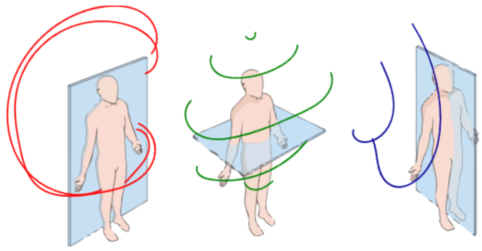

Figure 1: To calculate the reachable volume, the arm was swept through 9 predefined arcs in the frontal, transverse, and sagittal planes. These data were later combined to generate a 3D point cloud of fingertip positions.

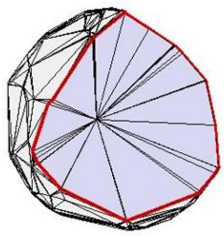

Figure 2: A convex hull surrounding all the 'finger' marker locations and the sternum marker was generated. The area connecting the extremes of the movement arcs behind the person's back was removed and replaced by a surface joining these extremes to the sternum to avoid overestimation of the workspace volume.

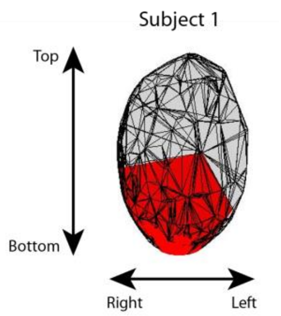

Figure 3: 3D reachable volume as viewed from the front. The combined volume shown in both grey and red is the unharnessed reachable volume, and the smaller red sub-volume is the harnessed reachable volume.

DISCUSSION

This study has introduced novel methods for evaluating reachable workspace and user control over prehensor aperture for a body-powered prosthesis. Clearly, an 'ideal' prosthesis would offer the user the ability to position and orient the prehensor at will within his/her unrestricted workspace, and to fully exploit the 'Mechanical aperture RoM' anywhere within this volume. The methods introduced here provide an objective approach to evaluating how far a given design is from this ideal.

Participants encountered major restrictions to both their reachable workspace and their ability to fully exploit the 'Mechanical aperture RoM' with their arm in different postures throughout the workspace. This is perhaps unsurprising and already recognized as an issue by clinicians who recommend a few different approaches to setting the cable length [5]-[7]. It is worth noting that the setup procedure used in this study (which was non-standard due to pilot work highlighting the infeasibility of employing standard approaches) resulted in a longer cable setups than the traditional approaches, and as such, these traditional approaches could result in an even greater reduction in reachable workspace and a greater number of positions where the user achieves <=50% of the full 'Mechanical aperture RoM'. These methods could be used to objectively evaluate alternative setups.

This proof-of-concept study offered a novel approach to the quantification of a body-powered prosthesis user's reachable workspace and their ability to exploit the 'mechanical aperture RoM' of the prehensor within that workspace. To interpret the results of our study and similar future studies there is a need to better understand the implications of a reduced reachable workspace and aperture control limitations for the user's daily life. The emerging field of real-world monitoring of prosthesis use [8], [9] may offer useful approaches which could be exploited here.

ACKNOWLEDGEMENTS

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Track: Myoelectric Control Algorithms

ACTION MYOELECTRIC CONTROL FOR ADVANCED HAND PROSTHESES VIA MULTI-LABEL CLASSIFICATION

Agamemnon Krasoulis, Kianoush Nazarpour

School of Engineering, Newcastle University

ABSTRACT

We propose action control, a novel approach for myoelectric independent digit control based on multi-label classification. At each time step, the decoder classifies movement for each controllable degree-of-freedom (DOF) into one of three categories: open, close or stall (i.e., no movement). The user employs continuous feedback information to estimate and minimise the mismatch between target and current digit positions. We implemented the proposed action controller and evaluated its real-time performance with 3 transradial amputee—two bilateral, one unilateral—, whilst they controlled a six-dimensional computer interface with surface electromyography (EMG) signals. We benchmarked the performance of the algorithm against the state-of-the-art in myoelectric digit control, that is, position control using multi-output regression. We found that action control consistently and substantially outperformed position control. Furthermore, all participants rated action higher than position control in a series of questions in a post-experimental survey and expressed and overall preference for the former. The proposed algorithm warrants further investigation in the future by transferring the control space from a computer display onto a real prosthesis and evaluating performance during activities of daily living.

INTRODUCTION

The holy grail of upper-limb myoelectric prostheses is individual control of digits in a continuous space [1]. Several teams have previously attempted to use regression algorithms to map electromyography (EMG) features onto digit positions/velocities offline [2-5]. Only a few studies, however, have demonstrated real-time digit position control in amputees [6-8]. Furthermore, the feasibility of using this paradigm to enable the user to perform object manipulation and activities of daily living in an unconstrained environment yet remains to be demonstrated.

We propose *action control*, a novel approach for individual digit control with EMG signals. In the heart of the control algorithm lies a multi-label classifier, which decodes movement intent for each controllable degree-of-freedom (DOF) into one of three classes: open, close or stall (i.e., no movement). We implement our proposed algorithm in real-time and evaluate its performance with three transradial (i.e., below-elbow) amputee participants using a six-dimensional control interface. We show that action control can systematically and substantially outperform the state-of-the-art for myoelectric digit control, which is based on position control via multi-output regression.

METHODS

Participant recruitment

We recruited three transradial amputee volunteers. Two of the participants had bilateral and one had unilateral amputation. Participant 2 performed two experimental sessions with alternate sides, thus the total number of sessions was $n = 4$. Experimental procedures were in accordance with the Declaration of Helsinki and approved by the local ethics committee at Newcastle University. Participants gave written informed consent prior to the experiments.

EMG recording system

We recorded surface EMG activity with 16 Delsys® Trigno™ sensors placed around the forearm in two rows of eight equidistant electrodes. Prior to sensor placement, we cleansed participants' skin using 70% isopropyl alcohol swabs. We visually inspected the quality of all EMG channels and used adhesive tape to secure sensor positions. The EMG sampling rate was fixed at 2 kHz.

Signal pre-processing and feature extraction

We processed EMG data using a sliding window with overlap. The length of the window was set to 128 ms and the overlap to 50%. Two features were extracted from each EMG channel, namely, waveform length and log-variance.

Prosthetic hand

We used the Robo-limb™ hand to demonstrate target postures to participants. The hand is similar to the Össur® i-Limb® Ultra hand and comprises six motors controlling thumb rotation and flexion/extension of all digits. The hand was powered by an external power supply unit (7.4 V/7 A) and operated by a laptop computer via a CAN bus connection.

Training data collection

We instructed participants to perform imaginary movements with their phantom limb, which were instructed on the prosthesis. The following single-digit and grip exercises were included: thumb opposition/reposition; thumb, index, middle, ring and little finger flexion/extension; cylindrical and lateral grip opening/closing. Participants performed 12 repetitions for each exercise and myoelectric data were recorded and stored on disk.

Control schemes and decoder training

During the interval between training data collection and real-time control, two types of decoders were trained: 1) a multi-output regression mapping EMG features onto digit positions (*position control*); and 2) a multi-label classifier decoding EMG features onto one of three classes: open, close or stall (i.e., no movement) (*action control*). In both cases, the target vector was six-dimensional, that is, the number of controllable DOFs.

Real-time control task

Participants were instructed to use their muscles to control a six-dimensional bar interface on a computer display. Prior to the start of the trial, the target posture was demonstrated on the prosthesis. Upon completion, a cue sound initiated the start of the *preparation phase* of the trial and six pairs of bars appeared on the screen. For each DOF, a fixed red bar indicated the target position and a blue bar showed the position that was controlled by the participant. Participants were given 5 s to match the blue bars to the red ones as closely as possible. A second cue sound initiated the start of the *evaluation phase* of the trial, which lasted for 1 s. Ten target postures were included, which comprised both single-digit and full-hand grip patterns: thumb opposition; thumb, index, middle, ring and little finger flexion; cylindrical, lateral and tripod grips; and index pointer. Note that not all exercises were included in the training set. Participants performed 10 blocks of trials for each control condition. Every target posture was included exactly once within each block in a pseudo-randomised order.

Evaluation

At the end of each trial, participants received a score characterising their performance during the evaluation part of the trial. The score was based on the median absolute error between the target and controlled positions and was normalised between 0% and 100%.

Post-experimental questionnaire

At the end of the experimental session, participants were asked to rate the two control schemes, namely, position and action control, based on the following three questions: 1) the interface was easy; 2) the interface was intuitive; 3) I found it easy to adapt to the interface. Ratings ranged from 1 (strongly disagree) to 5 (strongly agree) and half scores (e.g., 3.5) were also allowed. Participants were finally asked to indicate their overall preference. Participant 2 answered the questionnaire twice, once after each session, and respective scores were averaged.

Statistical analysis

For each participant, the target presentation order was the same for the two conditions (i.e., paired measurements). To compare performance between the two algorithms, we used two-sided Wilcoxon signed-rank tests with Holm-Bonferroni correction to account for multiple comparisons. The condition order was counter-balanced across participants.

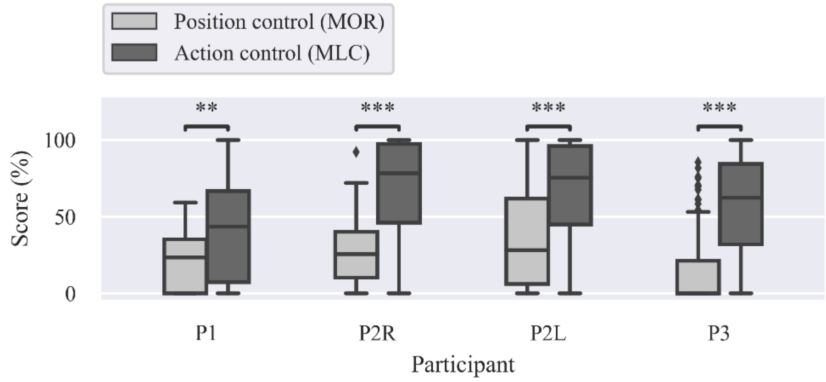

Figure 1: Performance comparison between position (i.e., multi-output regression) and action (i.e., multi-label classification) control. The performance score characterised the match between target and controlled positions during the evaluation phase of the trial. Higher values indicate better performance. Solid lines, medians; solid boxes, interquartile ranges; whiskers, overall ranges of non-outlier data; diamonds, outliers; double asterisk, $p < 0.01$; triple asterisk, $p < 0.001$.

Table 1: Post-experimental questionnaire

Range: 1 (strongly disagree) to 5 (strongly agree); PC, position control; AC, action control

Participant	**Question**						
	Interface was easy		**Interface was intuitive**		**I found it easy to adapt to the interface**		**Overall preference**
	PC	***AC***	***PC***	***AC***	***PC***	***AC***	
P1	2	4	1	4	3	5	AC
P2	2	3.5	3	4	2.5	3.5	AC
P3	3	4.5	2	4.5	2	4	AC

RESULTS

The performance results from the real-time control experiment are presented in Figure 1. The scores achieved by each participant with the two conditions (i.e., position and action control) are summarised using box plots. For all four sessions, action control (i.e., multi-label classification) significantly outperformed position control (i.e., multi-output regression). The differences in median performance were as follows: P1, $MD = 20.14$, $p < 10^{-2}$; P2R, $MD = 52.63$, $p < 10^{-13}$; P2L, $MD = 47.23$, $p < 10^{-10}$; P3, $MD = 62.32$, $p < 10^{-13}$.

The outcomes of the post-experimental questionnaire are presented in Table 1. All participants rated action higher than position control in all three questions. Furthermore, all three participants expressed an overall preference for action control.

DISCUSSION

We have introduced a novel paradigm for myoelectric digit control. At each time step, the algorithm decodes movement for each controllable DOF in one of three categories: open, close or stall. To reach a desired position, the user has to utilise the available feedback information—in our experiment visual from the computer display—to estimate the mismatch (i.e., error) between the target and current position(s) and take appropriate action(s) to minimise it. The controller can be viewed as an extreme, discretised case of velocity control; the velocity has a fixed value and is, thus, only parametrised by its direction. Using this approach, we can employ a multi-label classifier as the decoder, rather than a multi-output regression algorithm. One caveat of regression-based approaches is that noise in the input (i.e., EMG) space is propagated to the output, hence resulting in unstable control. To address this issue, it is common to smooth the output using a low-pass filter. Nevertheless, a large amount of smoothing is typically required to achieve a satisfactory outcome, which translates into a noticeable control delay. Classification, on the other hand, does not suffer from this limitation due to its discrete nature. Thus, by replacing the regression algorithm by a classifier we can achieve more stable digit control. Action control has an additional advantage. As opposed to position control, whereby a user has to hold a muscle contraction to retain a specific posture, with action control the user can completely relax once the target posture has been reached. This can result in more effortless control for the user.

We have previously shown that position and action control can yield comparable performance in a robotic hand tele-operation task with a data glove [9]. Here, we have provided a real-time myoelectric implementation of the two algorithms and have shown that action control can systematically outperform position control, which is considered as the state-of-the-art for prosthetic digit control. Moreover, all participants rated action higher than position control in a series of questions and expressed an overall preference for the former. As a future direction, we will compare the performance of the two algorithms using additional metrics. Finally, we will further evaluate action control by transferring the control space from a computer interface onto a real prosthesis.

CONCLUSION

We have proposed and evaluated a novel paradigm for myoelectric individual digit control based on multi-label classification. We have shown that it can systematically outperform the state-of-the-art position control approach based on multi-output regression. In the future, we shall further validate the algorithm by transferring the control space from a computer interface onto a real prosthesis.

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AN ALGORITHM CALIBRATED WITH CATEGORICALLY LABELLED EMG FOR END-TO-END ESTIMATION OF CONTINUOUS HAND KINEMATICS

Alexander E. Olsson[1], Nebojša Malešević[1], Anders Björkman[2], Christian Antfolk[1]

[1]*Dept. of Biomedical Engineering, Lund University*, [2]*Dept. of Hand Surgery, Lund University*

ABSTRACT

To restore limb functionality, control of a prosthetic hand should ideally be (I) proportional, i.e. produce speeds which varies in conjunction with changes in the latent intensity of muscle contractions, and (II) simultaneous, i.e. allow for both combined and independent steering of relevant kinematic degrees of freedom (DoFs). These desiderata are not straightforwardly attainable with classificatory pattern recognition applied to surface electromyography (sEMG), which only allows for the detection of a finite set of categorically encoded gestures. To alleviate such limitations, we here introduce a related approach for myocontrol which maps sEMG envelopes directly to multiple, continuously encoded DoFs, providing proportionality and simultaneity implicitly. The proposed method, termed myoelectric representation learning (MRL), is constituted by a deep learning topology and a domain-informed model training scheme. As with conventional pattern recognition, MRL operates on sEMG exclusively and is calibrated without ground truth limb kinetics, allowing for deployment with amputee users. We demonstrate the practical viability of MRL by implementing a virtual control interface driven by a setup consisting of 8 surface electrodes and capable of decoding 2 kinematic DoFs in real-time. Experiments with 10 healthy subjects, in which the interface was used to conduct tests yielding 5 numeric performance metrics, were performed to quantify the quality of myoelectric control afforded by MRL. Comparisons with the performance obtained from of a Linear Discriminant Analysis benchmark method on an identical test revealed that MRL outperforms the former in all computed measures of control efficacy.

INTRODUCTION

Pattern recognition applied to surface electromyography (sEMG) has for a time been considered a key component in the endeavour to make intuitively controlled, multiarticulate upper limb prostheses available to transradial amputees [1]. Despite countless reports of successful application of several variations of the technology in lab environments, widespread clinical adoption remains elusive [2]. Due to the notable level of reliability and stability required for practical viability, the few commercial implementations existing currently [3] make use of linear classification algorithms applied to a robust set of handcrafted signal features [4]. Within this *gesture detection* framework, speed of motion is typically modulated separately from classification by use of the mean average value of sEMG aggregated across all available channels [5]. Albeit functional and robust, this type of approach does not allow for true *simultaneity*, here defined as the ability to separately control multiple kinematic degrees of freedom (DoF) with mutually independent speeds.

This paper introduces an alternative method for intuitive, proportional, and simultaneous myoelectric control which functions via supervised machine learning and is constituted by (I) a computationally lightweight artificial neural network (ANN) and (II) an appertaining calibration strategy. Due to its reliance on kinematically influenced signal representations arising throughout the ANN model during use, the method is termed *Myoelectric Representation Learning* (MRL).

METHODS

10 able-bodied subjects (age range 26-49 years, 5 male and 5 female) participated in the current study, which consisted of two phases: acquisition of calibration data followed by evaluation of myocontrol efficacy. The study was approved by the Regional Ethical Review Board in Lund, Sweden and all subjects gave their written consent. Data acquisition and processing were performed with custom code written for and executed in Python 3.6. All hyperparameters were selected *ad-hoc* prior to the start of experiments via empirical work on subjects not part of the current study

Data Acquisition

sEMG signals were acquired with a Myo armband (Thalmic labs, Canada) consisting of 8 equiangularly spaced dry surface electrodes. At the start of each experiment session, the armband was placed enclosing the dominant forearm of the subject at a level approximately 1/3 of the distance from the humeroradial joint to the radiocarpal joint. sEMG signals were sampled at a rate of 200 Hz and were transferred at identical rate to a host desktop computer (on which all signal processing was performed) in real-time via Bluetooth. The subject was seated comfortably in a chair, approximately 1 m from the computer screen, with elbow resting on a table; the angle and position of the elbow could be varied freely by the subject at all times.

Table 1. The recorded calibration movements and their corresponding categorical target encodings.

Movement Class	Description	Ternary Encoding $\boldsymbol{y}$
0	*Rest*	[0, 0]
1	*Wrist flexion*	[-1, 0]
2	*Wrist extension*	[1, 0]
3	*Flexion of the digits*	[0, -1]
4	*Extension of the digits*	[0, 1]
5	*Wrist flexion and Flexion of the digits*	[-1, -1]
6	*Wrist flexion and extension of the digits*	[-1, 1]
7	*Wrist extension and flexion of the digits*	[1, -1]
8	*Wrist extension and extension of the digits*	[1, 1]

The current study entailed the decoding of two DoFs: (I) wrist flexion/extension and (II) flexion/extension of all digits simultaneously. Movement instruction stimuli were encoded with a ternary scheme, where each DoF could assume the values -1 (DoF active in one direction), 0 (DoF inactive), or 1 (DoF active in the opposite direction). All of the resulting 3^2=9 combinations possible in this framework (shown in table 1) were recorded. Prior to calibration data acquisition, subjects were instructed to perform each of the 8 nonrest movements classes with maximal voluntary contraction (MVC) for 5 seconds. This step served to familiarize the subject with the movement combinations under consideration and was furthermore used to compute an MVC magnitude value specific to each subject and movement by summing the mean absolute value over all 8 sEMG channels.

Calibration data was recorded by an acquisition program which prompted the subject to perform all nonrest movements for 3 repetitions, each lasting for a duration of 5 s and separated by 3 s of rest. To aid the subject in applying a sustainable and consistent level of contraction across movements, the mean absolute value of the sEMG signal, summed over all channels of a sliding window of length 0.5 s, was mapped to the height of a bar shown in real-time on the computer screen together with a threshold set to equal 50% of the movement-specific MVC magnitude computed earlier; subjects were instructed to keep the activity level as close to the threshold as possible. Once the program concluded, recorded sEMG was, together with the concurrent movement instruction stimuli information, saved and subsequently used for calibration of two different myoelectric control methods (an example of such calibration data is provided in fig. 1).

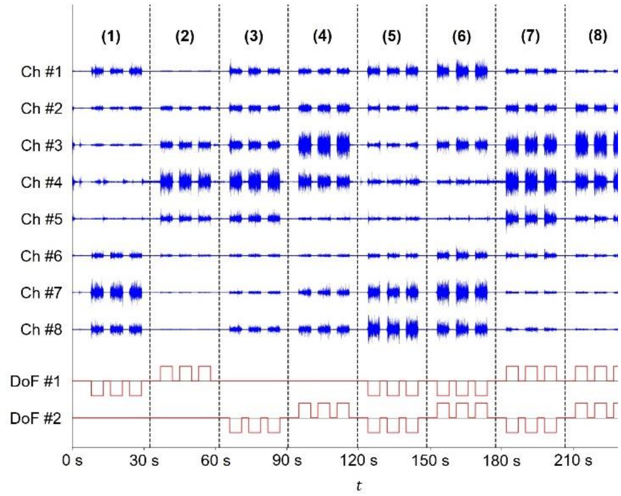

Figure 1. sEMG calibration data acquired from a single subject. **(1)** Wrist flexion. **(2)** Wrist extension. **(3)** Flexion of the digits. **(4)** Extension of the digits **(5)** Wrist flexion and flexion of the digits. **(6)** Wrist flexion and extension of the digits. **(7)** Wrist extension and flexion of the digits. **(8)** Wrist extension and extension of the digits.

Myoelectric Representation Learning

Before being applied for neural network training, the previously collected sEMG signals were subject to preprocessing in the form of an *envelope extraction* step followed by a *nonlinear rescaling* step. Envelope extraction entailed signal rectification and channel-wise lowpass digital filtering with a moving average filter of length 0.5 s (100 samples), yielding a nonnegative and unbounded signal matrix $\boldsymbol{E}^u$. Nonlinear rescaling entailed channel-wise linear rescaling, clipping and lastly transformation by the square root operator as is shown in equations 1 and 2 below.

$$E_{i,t}^{r} \leftarrow \frac{E_{i,t}^{u} - p_i^{1\%}}{p_i^{99\%} - p_i^{1\%}} \tag{1}$$

$$E_{i,t}^{Tr} \leftarrow \sqrt{max(0, min(1, E_{i,t}^{r}))} \tag{2}$$

$p_i^{1\%}$ and $p_i^{99\%}$ were the 1st and 99th percentile level, respectively, of the samples of the ith channel of $\boldsymbol{E}^u$. These preprocessing steps (I) guaranteed that all samples in $\boldsymbol{E}^{Tr}$, which were to be used for optimization, were constrained to the interval [0, 1] and (II) limited the impact of outlier sEMG samples on the resulting training data. The square root operator was included to bias resolution towards high levels of muscle contraction (i.e. $E_{i,t}^{r}$ close to 1). When the system later operated in real-time inference mode, input sEMG was identically processed using online filtering and the statistics $\boldsymbol{p}^{1\%}$ and $\boldsymbol{p}^{99\%}$ obtained from the calibration data.

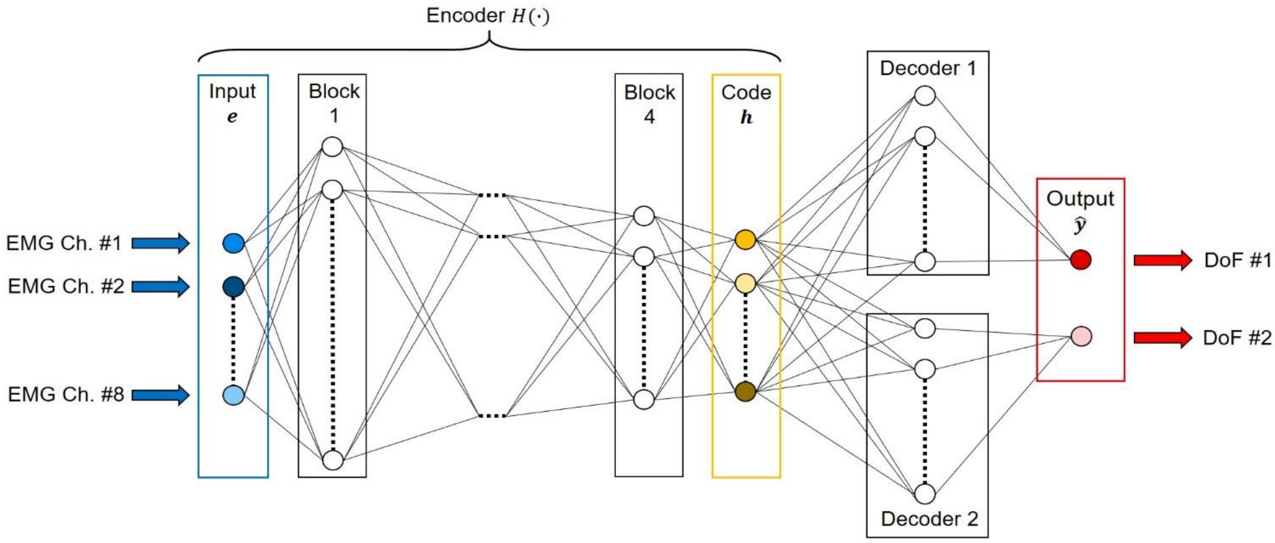

Figure 2. The regression neural network topology central to the presented approach.

The MRL topology, presented schematically in fig. 2, was composed of an *encoder subnetwork*, shared between the DoFs, and two separate *decoder subnetworks*, each specific to a DoF. The encoder network consisted of 5 fully connected *blocks*, each in turn consisting of a fully connected layer [6], a leaky ReLU activation [7], and layer normalization[8]. The number of output nodes for each encoder block was set to 128, 64, 32, 16, and 8, respectively, resulting in a *code size* of 8. Each decoder network operating on the generated signal representation consisted of one fully connected block of the same type, utilizing 128 hidden units, terminating in a fully connected layer with 1 linear output node, representing the inferred level of activity for one of the decodable DoFs.

Model training was performed via gradient descent with batch size of 4096 for 5000 iterations by the Adam algorithm [9] with $\eta = 10^{-4}$, $\beta_1 = 0.9$, $\beta_2 = 0.999$. The loss to be iteratively minimized was given by equation 3.

$$\mathcal{L} = \mathcal{L}_i + \alpha_c \mathcal{L}_c \tag{3}$$

α_c is a hyperparameter, set to equal 10^{-2}. $\mathcal{L}_i$ is referred to as the *inference loss* and given in equation 4.

$$\mathcal{L}_i = \|\hat{\boldsymbol{y}} - \boldsymbol{y}\|_1 \tag{4}$$

The regressand $\hat{\boldsymbol{y}}$ is the 2-element vector containing the DoF-wise continuous kinematics inferred by the ANN and $\boldsymbol{y}$ is the ground truth ternary encoding of the movement instruction stimuli concurrent with the sEMG envelope regressor sample. With $\mathcal{L}_i$ minimized, the ANN produces output which matches the movement intent of the subject.

$\mathcal{L}_c$ denotes the *contractive loss* and given in equation 5.

$$\mathcal{L}_c = \frac{1}{2 \cdot 8} \sum_{i}^{8} \sum_{j}^{2} \left(\frac{\partial \hat{y}_j}{\partial e_i} \right)^2 \tag{5}$$

The term $\frac{\partial \hat{y}_j}{\partial e_i}$ denotes the gradient of the jth output DoF with regards to the ith channel of the input sEMG envelope. With a minimized $\mathcal{L}_c$, ANN output will be sensitive to variations in the level of latent muscle activity (as proxied by the sEMG envelope), i.e. control will be proportional.

Benchmark Pattern Recognition Control

To verify the conjectured advantages of MRL, a benchmark proportional pattern recognition method for myocontrol based on linear discriminant analysis was implemented. All implementation details, including feature extraction, classifier architecture, and calculation of speed, were selected to be identical of those of Method 2 introduced by Scheme et al in [5]. The method is in its entirety henceforth referred to simply as LDA.

Quantitative Method Evaluation

A real-time virtual environment was implemented to quantify myocontrol efficacy for both MRL and LDA. To counteract confounding effect from acclimation, half of subjects were selected to evaluate MRL first whereas the other half were selected to evaluate LDA first (determined randomly). The output command of the evaluated method was mapped to the velocity of a *cursor* shown on the computer screen. Detection of wrist flexion/extension translated to cursor movements left/right, and detection of flexion/extension of digits translated to cursor movements down/up. In the test, subjects were instructed to steer the cursor towards a sequence of circular *targets*, generated at 20

Table 2. Summary of real-time performance metrics.

Name (abbreviation)	Description
Completion rate (CR)	The proportion of targets which were successfully reached.
Completion time (CT)	The average time elapsed between task start and completion
Path Efficiency (PE)	The average ratio between the straight-line distance from the starting point to the target and the actual distance traversed.
Overshoot (O)	The average number of occurrences wherein the cursor leaves the target prior to the end of the dwell time
Throughput (T)	The ratio $\frac{ID}{CT}$ between index of difficulty (ID) and completion time (CT), averaged across all successfully reached targets.

locations spanning all four quadrants with 2 radii, resulting in a set of 40 targets each covering either 0.6% or 2.3% of the total screen area. The order in which targets were presented was determined randomly for each subject. An index of difficulty ID was computed for each target as in [10]. As in earlier work [11], targets were considered successfully reached after a dwell time of 0.3 s and considered failed if not successfully reached within 20 s. The 5 performance metrics introduced by Williams and Kirsch in [10] (summarized in table 2) were computed for each subject and control method at the end of experiments.

RESULTS

Linear regression showed a strong relationship between ID and CT ($R^2 = 0.89$) for MRL across all subjects and targets. The corresponding value for LDA was computed as $R^2 = 0.81$, verifying the eligibility of the Fitts's law test and by extension the validity of the throughput metric. Aggregated performance metric summary statistics of both MRL and LDA from all subjects are presented in table 3.

CONCLUSIONS

The proposed algorithm (MRL) was found to be superior to conventional pattern recognition (LDA) in the sense of outperforming the latter in all computed measures of real-time efficacy of control. These results are encouraging, but need to be replicated with a larger subject sample size (ideally including amputee subjects) as well as have their stability over longer time spans be investigated.

Table 3. Means and standard deviations of metrics.

Metric	MRL	LDA
CR	99.25 ± 1.60	98.00 ± 2.45
CT	3.68 ± 1.14	5.25 ± 1.43
PE	55.33 ± 10.83	49.93 ± 7.90
O	0.53 ± 0.19	0.61 ± 0.26
T	0.67 ± 0.15	0.51 ± 0.13

The MRL model was successful in extracting kinematics pertaining to two separate DoFs, but required calibration data of every possible movement combination. For larger numbers of DoFs, the number of movement combinations grows geometrically, leading to infeasibly long calibration data acquisition phases. Notably, this drawback is not unique to MRL, but is shared by all contemporary pattern recognition frameworks aimed at multiarticulate myocontrol

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BRACHIOPLEXUS: MYOELECTRIC TRAINING SOFTWARE FOR CLINICAL AND RESEARCH APPLICATIONS

Michael R. Dawson[1,4], Heather E. Williams[2], Glyn S. Murgatroyd[3], Jacqueline S. Hebert[1,3], Patrick M. Pilarski[1,4]

[1]Division of Physical Medicine and Rehabilitation, University of Alberta, Canada; [2]Department of Biomedical Engineering, University of Alberta, Canada; [3]Glenrose Rehabilitation Hospital, Alberta Health Services, Canada; [4]Alberta Machine Intelligence Institute, Canada

ABSTRACT

Various control strategies are now available for myoelectric devices. The selection of the most appropriate strategy for an individual patient and training to improve their skills are important components to optimize user function with their myoelectric prosthesis. Existing myoelectric training software is often limited by not providing enough features to allow prosthesis users to try the multiple options for prosthetic hands, wrists, and elbows and the various control strategies used to modulate or switch between them. To address this gap, we developed an open-source training software for clinical and research applications called brachI/Oplexus that aims to provide a wider breadth of options and also be easy to use by non-technical users. The software supports several input devices (EMG systems), output devices (robotic arms), and methods for mapping between them (conventional and machine learning controllers). A comparison was performed between brachI/Oplexus and two commercial myoelectric software programs. Results from the testing showed that brachI/Oplexus had similar or slightly improved EMG signal separation and delay when compared to the commercial software. Several research labs and hospitals are already using this software, and by releasing it open source, we hope to lower the barrier of entry and encourage other clinicians and researchers to explore this area.

INTRODUCTION

Training prior to provision of a definitive device plays an important role during the fitting process for an upper limb myoelectric prosthesis. Not only does it allow a person with amputation to improve their skills at using the technology, but it allows them try various options for prosthetic devices and control strategies. One of the main limitations of the existing training systems is that they typically only allow for training of a single prosthetic device with a limited set of mapping options [1]. To allow a person with amputation to try the full breadth of devices and mapping options, it is necessary to borrow training systems from multiple manufacturers, which can be time consuming and logistically difficult. In order to solve this issue, we previously developed our own custom training system, called the Myoelectric Training Tool (MTT) [2], which included an electromyography (EMG) acquisition system, control software, and a desktop robotic arm with 5 degrees of freedom (DoFs) including shoulder rotation, elbow flexion/extension, wrist rotation, wrist flexion/extension, and hand open/close. This system has been successfully used for research at the University of Alberta since 2011 and for clinical training at the Glenrose Rehabilitation Hospital since 2015.

An upgraded version of the robotic arm, called the Bento Arm, was developed in 2013 [3,4] with stronger servos that included integrated sensors for measuring position, velocity, temperature, voltage and load. The Bento Arm also was designed to have a more anatomical appearance and could be mounted to a desktop stand or worn as a prosthesis by attaching it to a socket. In 2017, we released a compatible sensorized, multi-articulated hand, called the HANDi Hand [5,6] which included flexion of all digits as well as thumb rotation. Initial studies employing these devices used software based out of Robot Operating system (ROS) or MATLAB’s Simulink Realtime (SLRT) Operating System. While these software platforms worked well in a research environment we found they were difficult to translate into clinical environments, as they could take several hours to install on new computers and were not easy to operate by researchers or clinicians coming from non-technical backgrounds.

In order to improve the accessibility of our software and its applicability to both clinical and research environments, we developed a new version based on lessons learned from our prior research and clinical deployments. The main objectives of the new software were that it be easy to install and use, and allow for mapping between a wide array of input devices and robots. We also decided to make the software open source, to facilitate use by other research groups and hospitals.

Taking inspiration from the anatomical term ‘brachial plexus’, which is the main network of nerves that connects the brain and spinal cord to the arm, we named the software brachI/Oplexus (pronounced “brak-I-O-PLEX-us”), with the goal of providing an improved digital nerve center for connecting input devices to robotic arms.

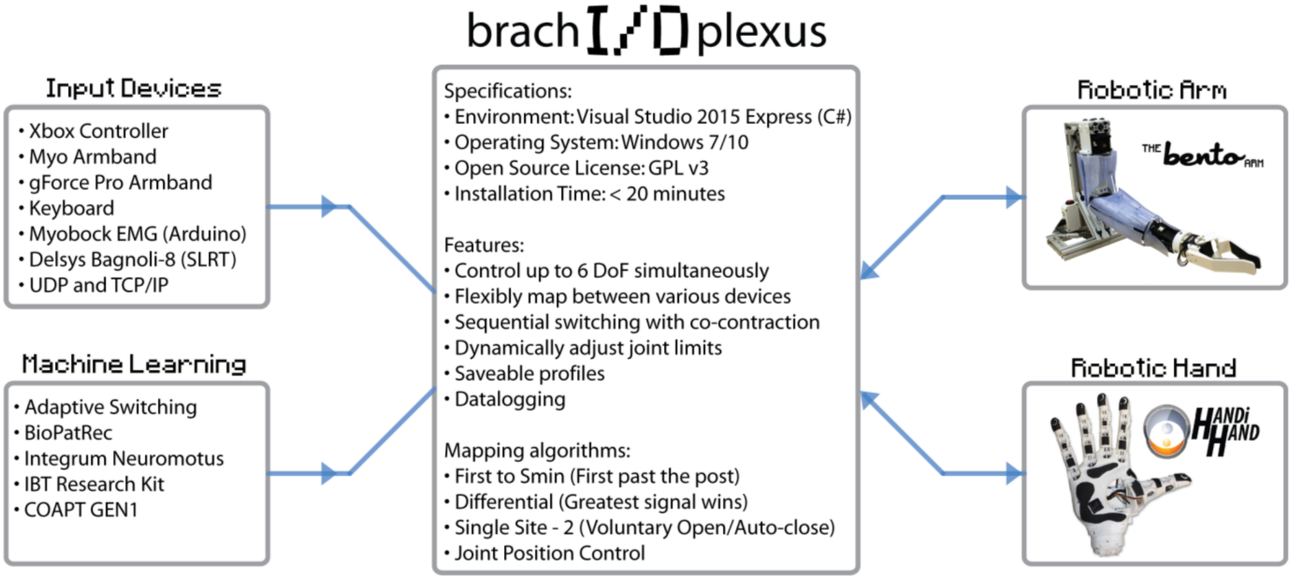

Figure 1: A high level software diagram showing how brachI/Oplexus takes in signals from input devices or external machine learning software and then maps them to joint positions or velocities on robotic devices.

SOFTWARE DESIGN

The main specifications and features of the brachI/Oplexus software are summarized in Figure 1. Initial requirements were determined by gathering feedback from clinical stakeholders, including clinicians, researchers, and persons with amputation. As part of this process, the developers of the software also gained valuable insight into possible improvements by directly observing over 50 myoelectric training sessions with clinicians and patients with amputation.

The development environment was selected based on criteria including timing performance, ease of use for both programmers and non-programmers, availability of libraries/interfaces, compatibility with existing software/hardware, and cost. Visual Studio 2015 Express using the C# language was chosen as it is free to use by researchers, students, and hobbyists and provides advanced tools for building graphical user interfaces and easy to use installer packages. Using the installer package combined with automatic driver installations from Windows Update, the entire software can be installed on a new computer and operational in less than 20 minutes. Initial performance testing in C# showed that we could achieve the desired step times of 200 Hz or faster for acquiring sensor data and sending motor commands to the Bento Arm.

To make brachI/Oplexus more accessible to external users, we released it open-source under a General Public License (GPL) v3 license, which allows the software to be freely used for both commercial and non-commercial purposes. As part of the open-source release, we developed extensive documentation explaining how to install and operate the software [7].

One of the key features of the software is the ability to control up to 6 DoFs simultaneously. This is especially useful when evaluating whether persons with transhumeral amputation are able to operate a multifunctional prosthesis. Additional DoFs that are not being evaluated for possible use in the definitive prosthesis, but that are useful for completing an engaging training task (i.e. shoulder rotation), can be controlled by the intact hand with the Xbox controller/keyboard or automated by the clinician.

Another key feature is the ability to flexibly adjust the mapping while the software is running without having to recompile the code or change settings in configuration files. All the required controls for selecting an input device, output device, and mapping algorithm are available through the graphical user interface as seen in Figure 2. Any combination of input device signals can be used to create multi-device mappings, and the signal settings including the minimum and maximum thresholds and gains are adjustable. The joint limits of the Bento Arm, including joint positions, velocities, and load can also be modified. These adjustments can be used to create movement envelopes to adjust the difficulty of tasks, and to improve the safety of the robot by limiting the torque and movement when in the same workspace as the operator (i.e. when the devices are being worn as prostheses). All of these settings are saveable as profiles, so that they can be easily reloaded at a later time. We find this feature is useful when seeing patients over multiple sessions in order to track their EMG settings and provide starting points for future sessions.

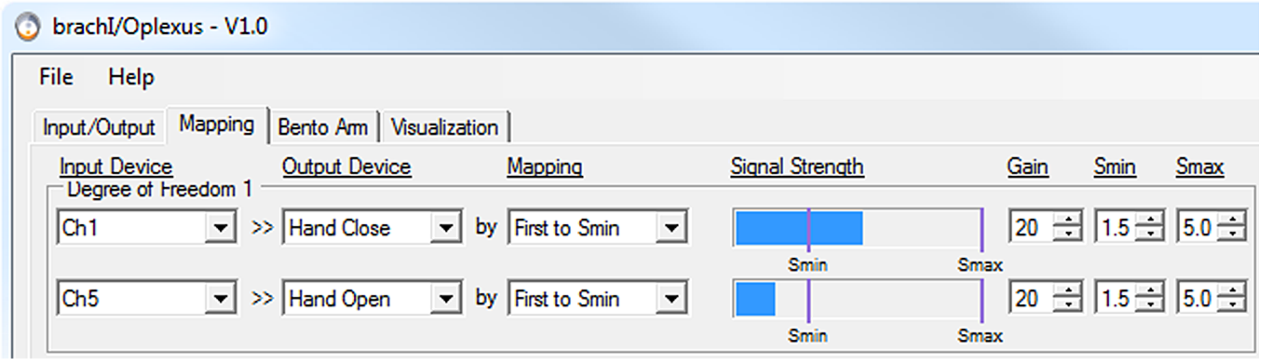

Figure 2: A screenshot of the mapping tab in brachI/Oplexus showing the adjustable settings for the 1st Degree of Freedom.

A data logging module was designed that can be used to log time series data from the input signals or feedback from the robots to a text file. In addition to specifying which signals to log, the sample rate can also be set to as fast as 200 Hz or as slow as desired. The data logging functionality can be used as part of research studies to record study variables and also as part of clinical training for logging training metrics.

The supported input devices are listed in Figure 1. Whenever possible existing open-source developer kits, libraries, and interfaces were used to connect to external hardware. The Microsoft Xbox controller and keyboard input devices are typically used for testing and demonstration purposes or as mentioned previously when controlling DoFs that are not yet available on commercial prostheses. Two different wireless armbands with EMG capabilities are supported including Thalmic Lab's Myo Armband (now discontinued, but still widely used) and Oymotion's gForce Pro Armband. We have found these armbands useful for demonstration purposes or for use with persons with transradial amputations. The Simulink Realtime interface (SLRT) is used to acquire signals from a Bagnoli-8 EMG Acquisition system with DE 3.1 electrodes (Delsys, Inc.) via a PCI 6259 data acquisition card (National Instruments, Inc.). The Arduino module is used to acquire signals from Myobock 13E200 electrodes (Ottobock, Inc.) via a custom designed wireless and battery-operated Arduino board. Both the SLRT and Arduino modules also allow for external buttons, FSRs, and EMG systems to be connected to the software. The UDP and TCP/IP network interfaces allow for external software to drive the arm. This is most commonly used for controlling the arm using machine learning software such as Adaptive Switching [8], BioPatRec [9], COAPT GEN1 (COAPT, LLC), IBT Research Kit (Infinite Biomedical, LLC), and Neuromotus (Integrum, Inc.). Several packet structures have been pre-designed to facilitate developing communication modules between brachI/Oplexus and new software. Another benefit of these network modules is that they can also communicate with Linux or macOS based operating systems.

The supported output devices include the Bento Arm and HANDi Hand which can be used together or separately in the software. For the Bento Arm, we communicate with the sensorized Dynamixels servos using the Dynamixel SDK, and for the HANDi Hand, we use an Arduino Mega to control its RC servos. There is also an option to directly control commercial myoelectric prehensors via the analog output on the Arduino or SLRT computer.

Several different mapping algorithms have been developed for brachI/Oplexus based on common control strategies available in commercial myoelectric software. Thus far, we have implemented control strategies for first to smin (also known as first pass the post), differential (also known as greatest signal wins), and single site-2 (voluntary open/automatic close). These are meant to serve not only as possible options for clinicians/patients using the software for clinical training, but also in research studies to provide a baseline to compare conventional controllers to machine learning controllers.

Also included in the software is a sequential switching module that allows for switching between up to 5 DoFs on the Bento Arm using a single pair of input signals. The switching can be triggered by exceeding a threshold with a dedicated signal or by co-contracting two signals together within a specified time frame. Several different feedback options are available to indicate to the user when they have switched and what DoFs they have switched to including visual, auditory, and vibratory feedback.

SOFTWARE COMPARISON RESULTS

We compared the signal processing and mapping algorithms in brachI/Oplexus to two commercial software programs. The comparison software included biosim v2.11.0.38 used with the i-Limb ultra revolution hand (Ossur, Inc.), and bebalance Version 3.5c used with the bebionic hand (Ottobock, Inc.). This study was approved by the Research Ethics Board of the University of Alberta (Pro00077893).

Table 1: EMG ratios and delays.

	Flexion agonist-to-antagonist	Extension agonist-to-antagonist	Delay (ms)
biosim + i-Limb + Myobock electrodes	4.0 ± 0.9	4.9 ± 1.4	151.7 ± 22.8
bebalance + bebionic + Myobock electrodes	3.2 ± 1.0	4.0 ± 0.6	150.1 ± 31.2
brachI/Oplexus + Bento Arm + Delsys Bagnoli electrodes	10.7 ± 5.9	12.8 ± 4.8	71.7 ± 13.2
brachI/Oplexus + Bento Arm + Myo Armband	6.3 ± 1.6	3.9 ± 0.6	142.6 ± 21.0

To compare the EMG performance across software, the rectified and averaged signals of 10 wrist flexion movements and 10 wrist extension movements were recorded from an able-bodied participant for each condition as specified in Table 1. The agonist-to-antagonist ratios were calculated for each of these movements, and average ratios and respective standard deviations are shown in Table 1. Larger agonist-antagonist ratios were observed with brachI/Oplexus with the Delsys Bagnoli electrodes, indicating that the signal separation might be a bit higher compared to the commercial myoelectric software.

To compare the delays observed with brachI/Oplexus to those observed with commercial devices, an iPhone 11 high speed camera (240 fps) was used to identify the time between when wrist extension was initiated and when the respective robotic hand began to open, for 10 repetitions. The average delays and respective standard deviations are shown in Table 1. The Bento Arm with brachI/Oplexus generally exhibited smaller delays than the i-Limb and bebionic hands, especially when paired with the Delsys Bagnoli electrodes.

Although there were some differences between brachI/Oplexus and the commercial software, in practise all of the systems behaved in a similar manner. If required, brachI/Oplexus also has the flexibility to mimic the commercial systems even more closely by adding a small delay to the motor commands or by using the Myobock electrodes with the Arduino module instead of the alternative EMG systems.

FUTURE WORK & CONCLUSIONS

Thus far, brachI/Oplexus has been used by 3 university research labs and 2 rehabilitation hospitals with other deployments ongoing. In the future we would like to add more input devices and output devices to the software to further improve its capabilities, so that it can be useful to even more clinicians or researchers. For example, we are in the process of creating a virtual reality version of the Bento Arm that would allow for training or research to take place in situations where it may not be practical to deploy physical robots (i.e. take-home training).

In conclusion, brachI/Oplexus is a fully functional myoelectric software with support for many different kinds of input devices, conventional and machine learning based controllers, and robotic arms. The wide array of features, improved accessibility, and dynamic adjustability make it well suited for clinical training and assessment as well as research applications.

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CLASSIFICATION OF TRANSIENT MYOELECTRIC SIGNALS FOR THE CONTROL OF MULTI-GRASP WRIST-HAND PROSTHESIS

Daniele D'Accolti[1,2], Andrea Mannini[1,2,3], Francesco Clemente[1,2], Itzel J. Rodriguez Martinez[1,2], Christian Cipriani[1,2]

[1]*The Biorobotics Institute, Scuola Superiore Sant'Anna, 56127 Pisa, Italy*
[2]*Department of Excellence in Robotics & AI, Scuola Superiore Sant'Anna, 56127 Pisa, Italy*
[3]*IRCC Fondazione Don Carlo Gnocchi, 50143 Firenze, Italy*

ABSTRACT

Decoding the neurophysiological signal generated by voluntary arm movements is one of the major challenges in rehabilitation engineering. The most investigated approach for hand prosthesis control is the continuous pattern recognition of myoelectric signals. However, this is based on the assumption that repeated muscular contractions produce consistent patterns of steady-state myoelectric signals. Notably, it is the initial, transient, phase of such signals that was shown to contain a deterministic structure. Here we investigated if both wrist and hand intended movements could be decoded from the transient phase of the myoelectric signal. Twelve healthy individuals performed one of four grasps and of five wrist movements simultaneously (20 combinations). Albeit the performance in recognizing both movements simultaneously was poor, the offline data analysis showed the feasibility of implementing a sequential wrist-hand embedded controller based on the transient phase.

INTRODUCTION

Individuals with a below-elbow amputation maintain part of the 18 extrinsic muscles that originally served the fingers and wrist. The electromyogram (EMG) recorded from these muscles can, in theory, be used to control a variety of motor functions in upper limb prostheses. Remarkably, the clinical state-of-the-art controller is still the two-state amplitude modulation controller proposed by Bottomley back in the '60s, [1]. In this controller, a single pair of agonist/antagonist muscles controls the opening and closing of the prosthetic hand. However, this scheme cannot differentiate between different muscular patterns pertaining to different hand movements, and, accordingly, cannot be used to control multiple grasps of a dexterous prosthesis intuitively.

An alternative approach is *pattern recognition,* as first proposed by Finley and Wirta in 1967, [2]. This technique is based on the premise that amputees can activate repeatable and distinct muscular contractions for each class of desired motion and that the associated EMG patterns can be identified and used to control the prosthesis accordingly. In this framework, Englehart and colleagues pioneered the development of *continuous* classifiers [3]–[5] that still represent the state of the art.

Remarkably, the assumption that repeated muscular contractions produce repeatable patterns of steady-state EMGs is weak. In fact, the steady-state EMG has very little temporal structure (it is mostly a random signal) due to the active modification of recruitment and firing patterns needed to sustain the contraction [6], [7]. For these reasons, time-averaged, compound statistical properties have to be extracted from the EMG signals before classification. To further improve the reliability of the latter, low pass filtering techniques (e.g. majority voting, velocity ramp or confident-based rejection) are usually applied to the output of the continuous classifiers [4], [8], [9].

While investigating the properties of the EMG at the onset of muscle contraction (the *transient*), Hudgins and colleagues observed a substantial degree of structure in the signals of upper arm muscles [10]. This observable structure was reported by others [11], and suggests a consistent orderly recruitment of motor units between contractions [7]. In our previous work, we exploited the transient EMGs generated during hand grasps/gestures (lateral, cylindrical, tri-digital grasp and hand open) to identify the intended movements using a simple representative classifier (i.e. the SVM). We demonstrated that the transients contained predictive information about the intended grasp, [12]. In this work, we investigated the possibility to extend the proposed method to the classification of both hand and wrist movements. We evaluated offline the performance of such a system in solving different classification problems, assessing its ability to operate with sequential or simultaneous wrist-hand movements. As the latter was not deemed sufficiently robust, the former was ported in a real-time system for a qualitative assessment.

MATERIAL AND METHODS

Twelve healthy subjects (age 26 ± 2.63 years old, 7 males, 10 right-handed) took part in the experiments after giving their informed consent.

Subject were asked to sit on a chair with the elbow flexed at 90 degrees on a table to limit the participant's fatigue during the test (Figure 1A). Eight EMG signals were sampled at 2 kHz (band-pass filtered at 10-900 Hz) using a signal amplifier (EMG-USB2+, OT Bioelettronica, Turin, Italy) and eight bipolar self-adhesive electrodes placed around the forearm (Figure 1B). In the described position, the subjects were asked to simultaneously perform one of the 20 possible combinations of two movements, involving: the hand (rest, lateral, tri-digital and cylindrical grasps) and the wrist (rest, flexion, extension, pronation and supination).

A custom-made graphical user interface was developed to help the subjects during the execution of the trials driving the type and timing of requested movements of both hand and wrist (Figure 1C). The interface also allowed the participant to pause the procedure in the interval between two movements to recover from fatigue, if required. Following the graphical hints in the interface, the participants were asked to: (i) execute a simultaneous movement of hand and wrist, (ii) keep the contraction for 3 seconds, (iii) move back to the initial resting condition. Three series of the 20 combinations were performed. Each series included five repetitions of each combination, for a total of (3 series × 5 repetitions × 20 combinations) 300 movements per participant. The order of movements was randomized among series.

The EMG signals were processed to extract the mean absolute value (MAV) on 100 ms windowed data, by sliding the observation window on a single sample basis. The

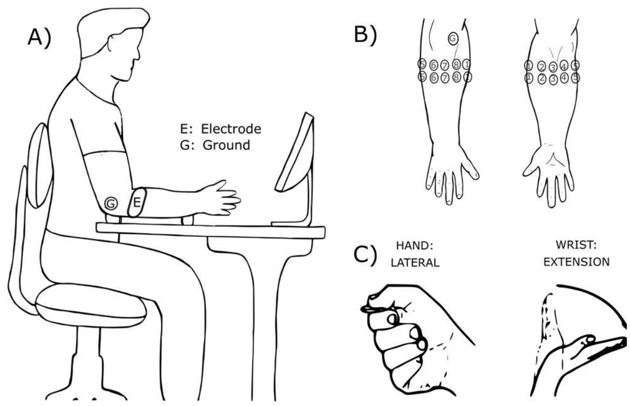

Figure 1: A) Experimental Setup. Participants were sitting in front of a monitor with the elbow flexed at 90 °.
B) Electrodes were uniformly distributed around the proximal part of the forearm. C) The Graphical User Interface informed the user on the next simultaneous movements to perform.

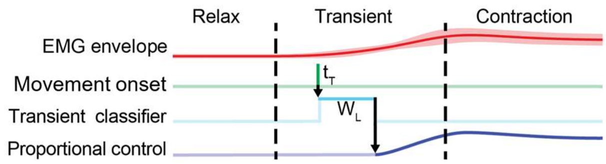

Figure 2: Transient EMG classifier concept. Once the transient detection algorithm (ODA) identifies an onset (at t_T), the transient window (W_L) is recorded and classified.

obtained signal was then down-sampled at 20 Hz and processed to extract the onset of muscle contraction through an onset detection algorithm (ODA). The ODA was applied to the derivative of the MAV. Specifically, for every class, the median peak of each series was calculated. Then, the minimum peak across series was set as the threshold.

In analogy with Kanitz et al. [12], after each detected onset, a different number of temporal MAV samples was extracted and provided to the classifier in order to establish which window length (W_L) allowed an optimal trade-off between classification accuracy and delay (Figure 2). Specifically, W_L ranged between 0 and 300 ms in steps of 50 ms (corresponding to 1,…,7 MAV samples). Using these features, a linear SVM classifier was trained and cross-validated for each subject, splitting the available data in 5 folds, assigned to each fold based on the order of repetitions of each series (leave-one-repetition-out approach). The classifier was tested in solving three different problems (P1-P3) with growing complexity:

P1. Recognizing grasps or wrist movements separately with two dedicated classifiers (four hand and five wrist classes).

P2. Recognizing grasps or wrist movements separately with one eight-class classifier.

P3. Recognizing grasps or wrist movements when performed simultaneously (20-class classifier).

A solution to P1 was searched to test if the results obtained in classifying the grasps [12] could be extended to wrist movements as well. Solving P2 would enable a sequential control of a robotic hand-wrist prosthesis. Finally, we also considered the more complex problem of recognizing hand and wrist movements performed simultaneously (P3).

Concerning the porting of the algorithm, an online classifier was implemented as suggested in Kanitz et al. [12].

RESULTS

The experimental recordings lasted for around one hour per participant, including the setup preparation. Results for all the addressed problems showed that the classification accuracy increases with W_L (Figure 3). This was expected as the longer the W_L, the more information is available to the

Table 1: Confusion matrix for the problem 2 for grasps and wrist movements (W_L = 200 ms)

Actual class	*Lateral*	*Pinch*	*Cylindrical*	*Extension*	*Flexion*	*Pronation*	*Supination*	*Rest*
Lateral	**142 (79.33%)**	10 (5.59%)	17 (9.50%)	3 (1.68%)	0 (0%)	0 (0%)	6 (3.35%)	1 (0.56%)
Pinch	5 (2.81%)	**140 (78.68%)**	5 (2.81%)	6 (3.37%)	4 (2.25%)	9 (5.06%)	9 (5.06%)	0 (0%)
Cylindrical	13 (7.22%)	2 (1.11%)	**147 (81.67%)**	1 (0.56%)	1 (0.56%)	7 (3.89%)	9 (5.00%)	0 (0%)
Extension	2 (1.11%)	2 (1.11%)	0 (0%)	**150 (83.33%)**	0 (0%)	6 (3.33%)	15 (8.33%)	5 (2.78%)
Flexion	2 (1.11%)	4 (2.22%)	1 (0.56%)	0 (0%)	**161 (89.44%)**	5 (2.78%)	6 (3.33%)	1 (0.56%)
Pronation	1 (0.56%)	3 (1.67%)	1 (0.56%)	8 (4.44%)	0 (0%)	**146 (81.11%)**	18 (10%)	3 (1.68%)
Supination	5 (2.81%)	1 (0.56%)	1 (0.56%)	5 (2.81%)	2 (1.12%)	11 (6.18%)	**151 (84.33%)**	2 (1.12%)
Rest	2 (1.11%)	0 (0%)	0 (0%)	0 (0%)	0 (0%)	0 (0%)	1 (0.56%)	**177 (98.33%)**
	Grasps Accuracy: 79.88%			*Wrist Accuracy: 84.68%*				
								Overall Accuracy 84.59%

classifier. However, W_L longer than 150 ms (or four MAV samples) improved the performance only slightly.

In general, the classification accuracy reached a plateau around W_L = 150 ms. Specifically, the performance did not improve significantly (Friedman test) for W_L > 150 ms for P1 and P2, and for W_L > 100 ms in the case of P3 (Figure 3). By comparing the different tested problems, accuracies for P3 were generally lower (58.86 % for W_L = 300 ms) than those obtained for P1 and P2 (93.33 % for W_L = 300 ms). Considering P2, the inclusion of wrist movements did not have a critical impact on the overall performance when compared to P1 (93.33 % vs 89.54 %, respectively). Specifically, wrist movements and grasps were classified with an overall accuracy of 84.68 % and 79.88 % (Table 1), respectively. In fact, wrist movements were classified more accurately than grasps (Table 1). This held true also for P1 and P3 (not shown).

Following the results mentioned above, the optimal solution was considered the one from problem P2. Thus, a single eight-class classifier was implemented and tested online. The outcomes from the online implementation and feasibility test are preliminary and qualitative in nature. Following a short training, consisting of 15 repetitions for each of the eight classes, the participant was able to use the online controller (supplementary video S1[1]).

DISCUSSION

To summarize, we claim that forearm EMGs patterns at the onset of a contraction contain predictive information about both upcoming hand and wrist movements. Moreover, this information can be used for real-time control of a wrist-hand prosthesis.

The transient EMG approach uses only the data contained in a short window associated to the onset of muscle contraction, which is known to contain a deterministic structure [10], [11]. The advantage of this approach is that classification is only necessary when a transient window is detected by the ODA, making the entire system less prone to errors. In addition, when errors occur, it is comparatively simple for the user to abort the ongoing grasp attempt and start anew. Importantly, since the contraction *precedes* the actual movement, the response time of the *transient* classifier is faster than that of a conventional continuous classifier.

Results from P1 complement the ones from our previous work [12] showing that the control strategy based on transients maintains very good performance also if applied to wrist movements (Figure 3).

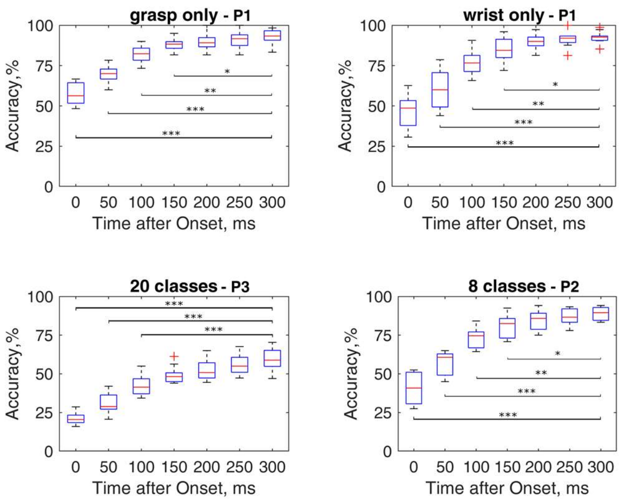

Figure 3: Results for considered problems as a function of the window length. The statistical analysis was performed with the Friedman test (*: 0.05 ≥ p > 0.01; **: 0.01 ≥ p > 0.001; ***: 0.001 ≥ p).

[1] https://drive.google.com/open?id=1WC2aWKbbIyQHhGwmHk0DMj1SfWQ02rIm

P3 represented the most challenging case, a 20-class classification problem involving simultaneous wrist and hand movements. The effort needed to acquire such a complex training set and the results obtained do not justify the use of a transient-based classifier for simultaneous hand-wrist control. The significant reduction in performance observed here with respect to P1 and P2 suggests that the information contained in the transients does not simply sum up constructively when more than a single anatomical district is involved in the movement. Several other groups also tried to investigate alternative methods for the simultaneous control of multiple degrees of freedom (DoF), but failed when the number of classes to be recognized increased above three or four [13], [14].

On the other hand, in P2 we analysed a standard eight-class classification problem that allows non-simultaneous hand-wrist movements. In this case, the performance were sufficiently good and only slightly worse than the ones obtained in P1. Notably, as analysed from Liu et al. [15], grasps and wrist movements are almost independent during normal reach-to-grasp tasks. In other words, a grasp is executed only after the wrist is already positioned. This perspective makes it feasible and natural to control the DoFs of a wrist-hand prosthesis in sequential manner. A result of these considerations is a reduction of control complexity.

This work has some limitations: (i) here we performed an offline analysis of the designed classifier and a qualitative evaluation of the online system (one subject case). It would be desirable to better evaluate the latter case, ideally including functional tests. (ii) We showed data acquired exclusively from healthy participants. An extension to amputee subjects is necessary to confirm the clinical usability of the algorithm. (iii) P3 would need a very extensive training phase (i.e. 15 repetitions × 20 classes = 300 trials) that is not compatible with a prosthetics application. We mitigated the problem with the continuous classifier (i.e. 15 repetitions × 8 classes = 120 trials), but the training phase is still quite demanding. Thus, the training phase part should be optimized to limit the number of repetitions needed to train each class. (iv) As a preliminary evaluation, we used a single feature: the MAV of the EMG. However, it is known that multiple time-domain features improve the accuracy of classification [16]. Future works will involve the introduction of new features, oriented particularly to an embedded real-time application.

Finally, we generalized the approach from our earlier work, extending the number of classes to include wrist movements. At the moment, a quantitative assessment of the real-time performance of a transient-based EMG controller are ongoing with both healthy and amputee subjects. Albeit we excluded the possibility to simultaneously control wrist-hand movements, we argue that a sequential control strategy based on the transient phase of the EMG could provide a natural and intuitive way to control a prosthetic device.

ACKNOWLEDGEMENTS

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A COMPARISON OF AMPUTEE AND ABLE-BODIED INTER-SUBJECT VARIABILITY IN MYOELECTRIC CONTROL

Evan Campbell, Jason Chang, Angkoon Phinyomark, and Erik Scheme
Institute of Biomedical Engineering, University of New Brunswick, Canada

Abstract—**Despite decades of research and development of pattern recognition approaches, the clinical usability of myoelectric-controlled prostheses is still limited. One of the main issues is the high inter-subject variability that necessitates long and frequent user-specific training. Cross-user models present an opportunity to improve clinical viability of myoelectric control systems by leveraging existing data to shorten training.However, due to the difficulty of obtaining large sets of data from amputee populations, data from intact-limbed subjects are often supplemented when building cross-user models; which may not translate well to clinical usability. In this preliminary study, the differences between intact-limbed and amputee cross-user electromyography (EMG) patterns were examined.Previously collected EMG data from 20 intact-limbed and 10 amputee subjects for different wrist, finger, and grasping gestures were analysed. Results using unsupervised clustering showed that amputees were consistently grouped into a different cluster than intact-limbed subjects and that additional clustering into more subgroups found larger differences between amputees than able-bodied subjects. Furthermore, a simple linear classifier was able to discriminate between able-bodied and amputee subjects using EMG from multiple gestures with 90% accuracy.These results suggest that using able-bodied subject data alone may be insufficient to capture the necessary inter-subject variance when designing cross-user myoelectric control systems for prosthesis control.**

I. Introduction

Although many applications of myoelectric control have been proposed in the literature since the 1990s, prosthesis control may still be considered as the predominant, and only commercial, application [1]. Nevertheless, despite many laboratory-based advances in pattern recognition-based myoelectric control (¿90% classification accuracy) [1], myoelectric-controlled prostheses still make a relatively limited clinical and commercial impact (e.g., only a quarter of patients with upper extremity amputations chose to use a myoelectric prosthesis [2]). This may be due to a gap between the academic state-of-the-art in myoelectric control and industry, which has been acknowledged and highlighted within the academic community [3]–[5]. One major limitation is high inter-subject variability, which limits the generalization of findings and necessitates frequent user-specific training and custom calibration [6], [7].

The main assumption of pattern recognition-based myoelectric control is that different types of muscle contractions exhibit distinguishable and repeatable signal patterns. Although distinguishable activation patterns are routinely found within a single user, there remain large differences between subjects. Most research studies, therefore, have adopted single-user (or subject-dependent) classification models, i.e., every user must train a system before his/her gestures can be recognized [1]. Few studies have investigated cross-user (or subject-independent) models and results have shown a marked decrease from the state-of-the-art (from ¿90% to 40%-60%) [8], [9]. Moreover, due to difficulties with access to persons with upper extremity limb deficiencies, most research studies have developed and investigated pattern recognition-based myoelectric control systems using intact-limbed subjects. Although relatively consistent algorithmic trends exist between the intact-limbed and amputee populations, an overall decrease in performance has typically been reported for the latter [5], [10].

In order to facilitate the development of cross-user models, particularly for clinical applications of myoelectric control, more information about subject-related differences in electromyography (EMG) patterns is required. The purpose of this preliminary study was, therefore, to examine the differences in surface EMG patterns between intact-limbed and amputee subjects across a large set of hand and finger gestures. Results are explored using data visualization and cluster analysis techniques.

II. Methods

A. EMG Data and Pre-Processing

Surface EMG data used in this study were taken from two NinaPro (Non-Invasive Adaptive Prosthetics) databases (3 and 7) [11], [12], which include data acquired from 20 intact-limbed subjects and 10 trans-radial amputated subjects. All subjects provided informed consent, and secondary consent was obtained for use of the dataset in this study. Additional details about the nature of the amputee subject data are shown in Table 1.

In these data sets, subjects performed a series of motions, including various individual-finger, hand, wrist, grasping, and functional movements. Databases 3 and 7 contain 52 and 40 total gestures, respectively, but the 38 common motions between the two databases were used for the present study. Each motion lasted 5 s, interrupted by 3-s rest time, and was repeated six times. Surface EMG data were collected using twelve Delsys Trigno Wireless electrodes; eight electrodes were equally spaced around the forearm (at the height of the radio-humeral joint), two electrodes were placed on the flexor and extensor digitorum superficialis muscles, and the remaining two electrodes were placed on the biceps and triceps brachii muscles. The sampling frequency was set to 2000 Hz. The data were cleaned of 50 Hz (and its harmonics) power-line interference using a Hampel filter. Erroneous movement labels were corrected by applying a generalized likelihood ratio algorithm [11].

TABLE I: Clinical characteristics of the amputee subjects (A1 and A2 from NinaPro Database 7 and A3-A10 from NinaPro Database 3). 'n/a' denotes data not available.

Subject	Amputated Hand	Years Since Amputation	Remaining Forearm (%)	Cause of Amputation
A1	Right	6	n/a	Accident
A2	Right	18	n/a	Cancer
A3	Left	6	70	Accident
A4	Right	5	30	Accident
A5	Right & Left	1	40	Accident
A6	Right	7	0	Accident
A7	Right	5	50	Accident
A8	Right	14	90	Accident
A9	Right	2	50	Accident
A10	Right	5	90	Cancer

III. Processing and Evaluation

The pre-processed EMG data were segmented for feature extraction using a window size of 200 ms and an increment of 100 ms. The commonly used Hudgins' time domain features [13]; mean absolute value (MAV), waveform length (WL), zero crossing (ZC), and slope sign change (SSC), were extracted from each window. A feature vector was then created from a series of the overlapped windows for further analyses.

Hierarchical cluster analysis (HCA) was used to create a dendrogram that identified homogeneous myoelectric patterns across the entire participant group (30 subjects). Briefly, HCA builds a hierarchical tree by combining a pair of clusters that leads to the minimum increase in total within-cluster variance after merging (Ward's criterion [14]), where the increase is a weighted squared Euclidean distance between cluster centers. Subjects in the same group have higher similarity (on average across 38 gestures, 12 muscles, and 6 repetitions) than the subjects in the other groups. Clusters in the data are determined by considering the height (or the distance) of each link in the cluster tree compared to the heights of the lower level links in the tree. If a link has a small increase in the height relative to the links below, it means that there are less distinct patterns differentiating the subjects joined at that level. Conversely, if a link height significantly differs from the links below, it means that there are more distinct patterns between them. This measure is referred to as the inconsistency coefficient.

Data visualization using principal component analysis (PCA), a commonly used feature projection method, was performed to better understand these complex myoelectric patterns. The main purpose of PCA is to summarize the important variance information in the data into the first few principal components (PCs), to facilitate visualization of distance and relatedness between populations in a reduced dimension. The identified PCs are linear combinations of the original features that can be used to express the data in a reduced form.

Finally, classification accuracies were computed using a linear discriminant analysis (LDA) classifier and a leave-one-out cross-validation technique to measure the performance of classification models in discriminating between gestures and between subjects. For gesture recognition, six clinically relevant motions were evaluated: wrist flexion, wrist extension, forearm pronation, forearm supination, power grip, and pinch grip. Classical within-subject gesture recognition was performed using leave-one-repetition-out cross-validation. For subject recognition, overall signal patterns were used (combining features from all repetitions of motions) in a leave-one-subject-out cross-validation approach. The goal of this task was to evaluate whether data could be classified as being from an able-bodied or amputee subject. This classification task was also repeated using each individual 200ms window of EMG data, again in a leave-one-subject-out cross-validation.

IV. Results

To validate previously reported results for intact-limbed and amputee subjects, the conventional gesture classification performance was computed for each group (Fig. 1). In keeping with previous findings, classification accuracies for the group of 20 intact-limbed subjects were significantly higher than the group 10 amputees (90.54% $\pm$ 3.6% $>$ 80.58% $\pm$ 9.8%; $p < 0.01$).

The results of the subject cluster analysis are shown in Fig. 2. It can be seen that the difference between the height of the links that connect the clusters (amputee and intact-limbed groups) and the mean height of the two links directly below is largest. In addition, the differences between the height of the links decreased as the number of clusters increased, and a plateau was found after six clusters were created. Thus, in this study, the two-cluster and the six-cluster solutions were employed.

When partitioning into two clusters (at the leftmost vertical dotted line in Fig. 2), Cluster 1 was found to consist purely of the amputee subjects (A1-A10) and Cluster 2, of purely intact-limbed subjects (S1-S20). When partitioning into six clusters (at the rightmost vertical line in Fig. 2), the previous clusters were retained, but were further subdivided. Cluster 1 was partitioned into 4 subgroups, with 1 subject in Cluster 1A, 4 subjects in Cluster 1B, 1 subject in Cluster 1C, and 4 subjects in Cluster 1D. The previous Cluster 2 was partitioned into 2 subgroups, with 10 subjects in Cluster 2A and 10 subjects in Cluster 2B.

Fig. 3 shows the projection of all subjects into PCA space. Two distinct clusters of patterns can be seen, highlighting the differences between intact-limbed and amputee subjects. A classification accuracy of 90% was found when using a simple LDA classifier to classifier whether the data from a given subject was able-bodied or amputee based on their

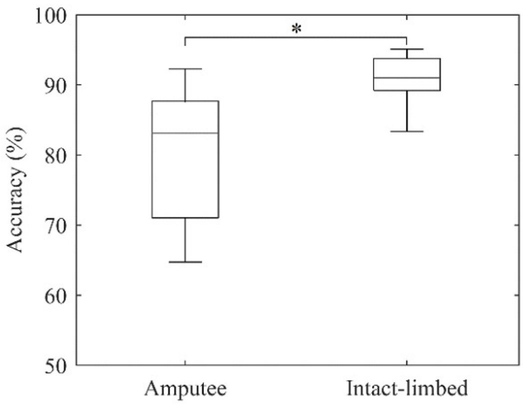

Fig. 1: Figure 1: Box plot of gesture classification accuracies using an LDA classifier with Hudgins' time domain features for amputee and intact-limbed subjects. * indicates significant difference ($p < 0.01$).

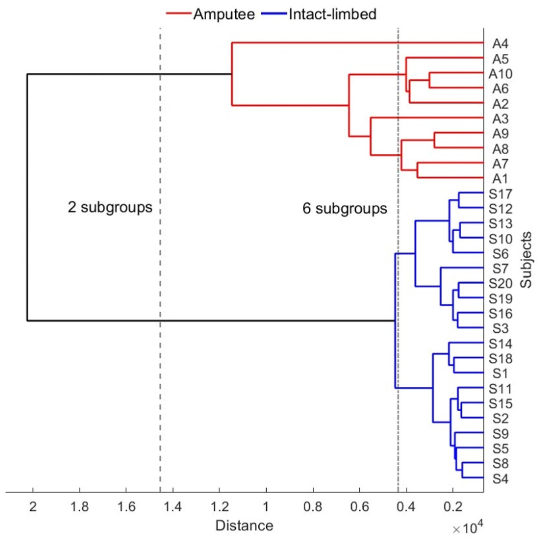

Fig. 2: Ward's linkage dendrogram of the hierarchical clustering of the overall myoelectric patterns representing the two-group and the six-group solutions. Participant numbers are indicated.

overall signal patterns. Although overall signal patterns were distinct, no differences were observable between the groups even when classifying a single frame of EMG as being from an able-bodied or amputee subject. A mean accuracy of 66% (min: 46%, max: 78%, chance: 66%) was observed across all subjects and motions classes.

V. Discussion

The main purpose of this study was to determine whether myoelectric patterns for intact-limbed and amputee subjects could be classified into homogeneous subgroups. The HCA approach was successful in identifying two distinct subgroups (yielding the highest inconsistency coefficient value: 4.38) based on overall myoelectric patterns. Although it would be expected that there are differences between intact-limbed and amputee subjects, it is quite surprising that an unsupervised learning algorithm could create two subgroups that discriminate myoelectric patterns of amputees and intact- limbed

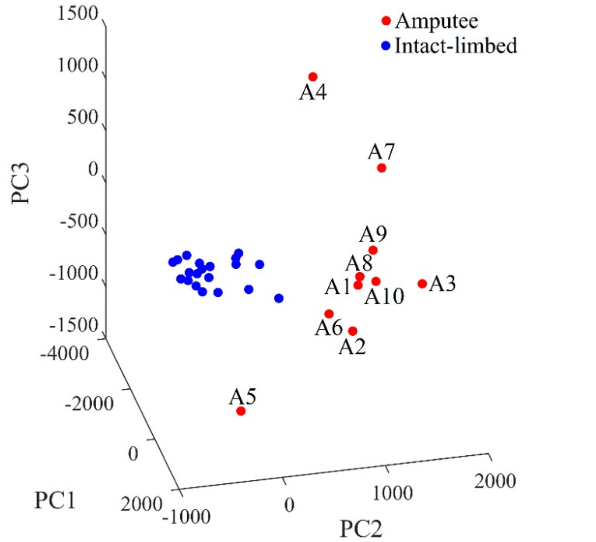

Fig. 3: Scatter plot of the first three PCs representing overall myoelectric patterns for 10 amputees (red dot) and 20 intact-limbed subjects (blue dot). The first three PCs explained 54% of the total variance.

subjects nearly perfectly (Fig. 2 and 3). From observation of Fig. 3, it appears as though a non-linear classifier could achieve 100% classification using only 2-3 PCs. Campbell et al. [10] investigated the differences between amputees and intact-limbed subjects using 58 state-of-the-art myoelectric features and suggested that most features in both time domain and frequency domain extract the same information for both subject groups. However, the migration of several amputee EMG features was found and can partially explain the performance degradation in amputee subjects (Fig. 1) (i.e., less information content is extracted using some EMG features for amputees). These findings suggest that when access to amputee populations is limited and able-bodied data is supplemented, outcomes of investigations on EMG features, dimensionality reduction, and classification algorithms should expect performance degradation when translating back to amputee populations. If a research study would like to develop a cross-user or subject-independent classification model for myoelectric-controlled prostheses, EMG data from amputee subjects is likely necessary given the noticeable difference in their patterns as compared to their intact-limbed counterparts (Fig. 1 and 3).

When 3-5 clusters were formed in Fig. 2, one group consisting of all the intact-limbed subjects remained consistent while the amputee group was partitioned into subgroups. This finding suggests that inter-subject variability in the amputee population is higher than between able-bodied subjects. A higher standard deviation of the classification accuracies for amputees (Fig. 1) also supported a higher inter-subject variability in amputee population. When the number of clusters was increased to six (yielding the second highest inconsistency coefficient value: 3.11), intact-limbed subjects were also divided into two subgroups. Some interesting characteristics of the six subgroups of subjects were found. For the two able-bodied subgroups, Cluster 2A provided slightly higher feature values compared to Cluster 2B. Cluster 1A, which contained only subject A4, provided the highest values for amplitude-based features (MAV and WL) among all subgroups but provided the lowest values for the complexity and frequency information-based features (ZC and SSC). It should be noted that the variance of feature values for this subject was very high, which could be due to noise or poor contraction repeatability.

Cluster 1D, which consists of 4 amputee subjects, provided the lowest values for the amplitude-based features, but the highest values for the complexity and frequency information-based features. It should be noted that most subjects in this group had prior experience in using a myoelectric prosthesis, suggesting that learning may play a role in cross-user differences. Cluster C1 consisted of only subject A3, the only subject with a left amputated hand and using a cosmetic prosthesis. Both subjects with an amputation due to cancer, were clustered together, in Cluster 1B. No meaningful trends were found for other clinical characteristics such as years since amputation, the remaining forearm percentage, degree of phantom limb sensation, and DASH (disability of the arm, shoulder and hand) score.

Overall, these findings suggest that the adoption of data from able-bodied subjects for the investigation of EMG features, dimensionality reduction, and classification algorithms, should be done with caution when focused on clinical applications for amputees. Specifically, even unsupervised clustering methods identified two distinct groups of subjects: one with all amputees and the other with all intact-limbed subjects. Of the subgroups, the amputee subgroup demonstrated much higher inter-subject variability. These results suggest that EMG data from amputee subjects is necessary for creating cross-user myoelectric-controlled prostheses, as their myoelectric patterns are considerably different than their intact-limbed counterparts.

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DIFFERENCES IN PERSPECTIVE ON INERTIAL MEASUREMENT UNIT SENSOR INTEGRATION IN MYOELECTRIC CONTROL

Evan Campbell, Angkoon Phinyomark, and Erik Scheme
Institute of Biomedical Engineering, University of New Brunswick, Canada

ABSTRACT

Recent human computer-interaction (HCI) studies using electromyography (EMG) and inertial measurement units (IMUs) for upper-limb gesture recognition have claimed that inertial measurements alone result in higher classification accuracy than EMG. In biomedical research such as in prosthesis control, however, EMG remains the gold standard for providing gesture specific information, exceeding the performance of IMUs alone. This study, therefore, presents a preliminary investigation of these conflicting claims between these converging research fields. Previous claims from both fields were verified within this study using publicly available datasets. The conflicting claims were found to stem from differences in terminology and experimental design. Specifically, HCI studies were found to exploit positional variation to increase separation between similar hand gestures. Conversely, in clinical applications such as prosthetics, position invariant gestures are preferred. This work therefore suggests that future studies explicitly outline experimental approaches to better differentiate between gesture recognition approaches.

INTRODUCTION

Gesture recognition using electromyography (EMG) pattern recognition has a long history of use in biomedical and clinical applications, such as myoelectric control of prosthetic devices and other assistive or rehabilitative technologies. These devices leverage residual motor function to enhance quality of life limited by neurological (stroke [1]) or physical impairment (amputation [2]). The emerging interest in hand gesture recognition as a general human-computer interface (HCI) for consumer applications, such as virtual reality, has large commercial incentives and has therefore accelerated in recent years. The use of wrist- or forearm-worn EMG devices combined with inertial sensors (i.e., accelerometer (ACC), magnetometer (MAG), or gyroscope (GYR)) have demonstrated the potential of such gesture recognition interfaces during offline classification studies [3]. These multi-modal devices have been validated in both biomedical and general HCI studies; however, the conditions of gesture elicitation differ between the two applications.

Biomedical applications of EMG pattern recognition typically require accurate recognition of physiologically appropriate gestures that are robust to variability of daily-living; simply put, the gestures should be reliably decoded regardless of limb posture and contraction intensity, among other factors [4]. Limb posture and contraction intensity variability degrades the usability of clinical EMG pattern recognition systems meaningfully, as gesture recognition accuracies were found to decrease on the order of 13% and 20% for these factors, respectively, across several studies [5]. Interventions in the form of training strategies [6], algorithmic solutions [7], or multi-sensor approaches [8] have lessened this degradation and led to more reliable use of myoelectric control. Multi-sensor approaches using EMG and ACC measurements from many positions have altogether removed degradation caused by static limb positions in recorded positions by sequential use of a position-classifier using ACC, followed by a position-specific EMG classifier for gesture recognition [8]. No application other than position recognition, however, has been validated for non-mechanomyographic ACC measurements within clinical EMG pattern recognition studies.

Alternatively, general HCI applications of EMG pattern recognition desire accurate recognition of distinct gestures; the gestures in these application are no longer required to be invariant to daily-living variability and may selectively harness position variability to become more distinct. Consequently, inertial sensors have been found to outperform EMG sensors in terms of gesture recognition accuracy [3], [9]–[11]. For instance, gesture recognition using MAG achieved 93% accuracy across 40 motion classes, whereas EMG achieved only 65%. The different interpretation of the application and value of inertial measurements between biomedical and HCI studies is a current area of confusion in the field that warrants further clarification.

This paper aims to highlight the main differences between biomedical and HCI studies of EMG pattern recognition by examining the differences between gesture elicitation studies. Specifically, this study focused on the differences in the gestures performed and the differences in the use of inertial information. Differences in gestures are presented through visualization of signals, whereas the differing use of inertial information is presented through classification outcomes using EMG and ACC feature sets.

METHODS

Datasets

Two public datasets were adopted to represent biomedical and HCI gesture recognition studies; the Fougner [8] and NinaPro7 [9] datasets, respectively. All subjects provided informed consent, and secondary consent was obtained for borrowed datasets. The biomedical dataset was collected using 8 bipolar Ag/AgCl electrodes (EMG) and 2 tri-axis accelerometers. Twelve intact-limbed subjects performed 6 motions (wrist flexion, wrist extension, wrist pronation, wrist supination, hand

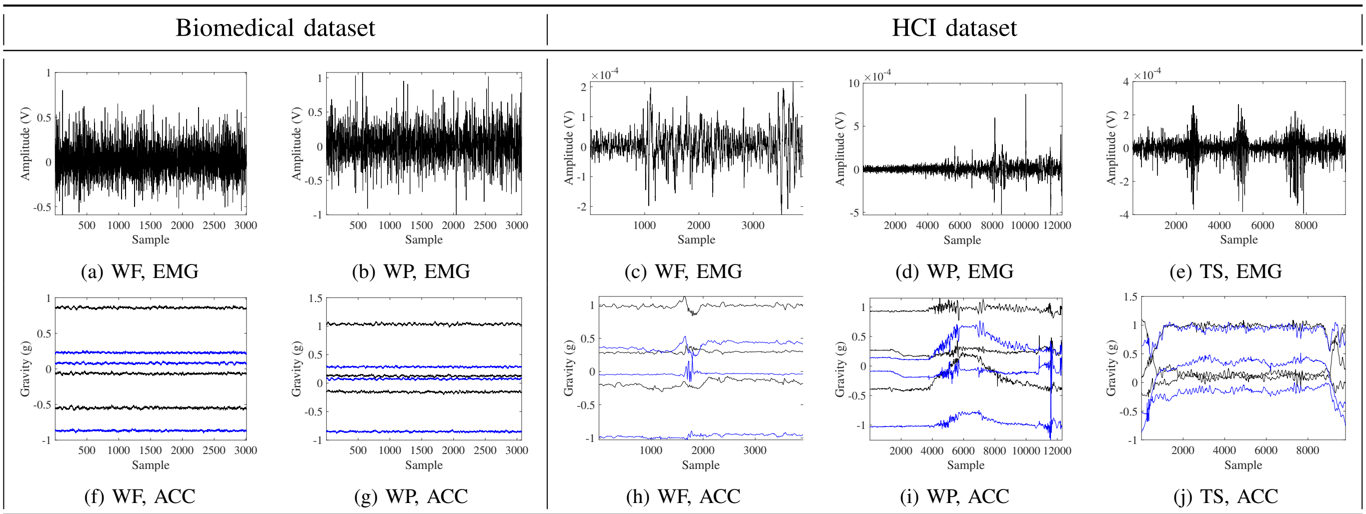

Fig. 1: EMG and ACC measurements (unfiltered) from the biomedical and HCI datasets. The first row contains the EMG elicited during wrist flexion (WF), wrist pronation (WP), and a turning screw (TS) gesture. The second row contains the accelerometer readings for the same contractions, where the black lines represent the x, y, and z components of a forearm mounted sensors and the blue lines represent measurements simultaneously taken at the biceps.

close, and pinch grip) and no motion, where each motion was repeated 10 times in 5 different static limb positions. The HCI dataset contained 12 bipolar Ag/AgCl electrodes and 12 tri-axis accelerometers. Twenty intact-limbed subjects performed 40 dynamic gestures (8 finger gestures, 9 wrist gestures, and 23 grasping gestures), where each motion was repeated 6 times with limb position unspecified. The gestures of the HCI dataset were segmented into 3 gesture sets: HCI-A, a set matching the biomedical dataset gestures, HCI-B, a subset containing 8 finger gestures, and HCI-C, a subset containing 23 grasp gestures. A sample of EMG and ACC signals from both datasets is given in Fig. 1.

Data preparation

The EMG signals from both datasets were pre-processed by a 60 Hz or 50 Hz notch filter and 20-450 Hz bandpass filter to remove power-line interference and motion artefacts, respectively. The ACC signals were pre-processed using 1 Hz low-pass filters, to remove accompanying sensor noise from measurements. Both EMG and ACC signals of all channels were segmented into overlapping windows using window length and increment of 200 and 100 ms, respectively.

Features were extracted from each window to create 2 EMG and 2 ACC feature sets. The EMG feature sets were the Hudgins' time-domain (TD) feature set [12] (mean absolute value, zero crossings, slope sign change, and waveform length), and the time domain power spectral descriptors (TDPSD) feature set [7]. The ACC feature sets were the median feature set (MED) and root mean square (RMS) feature set.

Classification problems

The four feature sets of all four datasets (biomedical, HCI-A, HCI-B, and HCI-C) were used in three classification tasks, where applicable, to validate claims proposed by previous studies.

1) Multi-gesture position classification: Classifiers were trained with feature vectors from all gestures with the class label selected as the position of the gesture. Only the biomedical dataset was used for this analysis, as the HCI dataset did not specify any specific limb positions.
2) Within-position gesture classification: Classifiers were trained with feature vectors from an individual position with the class label being the associated gesture. This process was repeated for all positions in the case of the biomedical dataset and only a single position was assumed for the HCI dataset.
3) Sequential classification: Classifiers were first trained following the multi-gesture position classification task, where feature vectors were used to predict position. Subsequently, the position was used to select the appropriate position-specific gesture classifier, as was conducted in the within-position gesture classification task. As the HCI datasets did not provide labelled positions, they were excluded from this task.

All classification tasks were performed using within-subject leave-one-trial-out cross-validation using linear discriminant analysis (LDA), quadratic discriminant analysis (QDA), k-nearest neighbours (kNN, k=5), and random forest (RF, 10 trees) classifiers. Accuracies are presented as mean + standard deviation, where the mean accuracy is the mean accuracy across all subjects and cross-validations, and the standard deviation is the standard deviation across subjects.

RESULTS

The multi-gesture position recognition results using ACC MED, ACC RMS, EMG TD, and EMG TDPSD feature sets are shown in Table I for the biomedical dataset. The within-position gesture recognition results of the biomedical and HCI datasets were presented in Table II. The LDA classifier was found to have the best performance among classifiers for all datasets in this latter classification task, again justifying

TABLE I: Multi-gesture position recognition accuracy (mean+std of subjects) across positions of the biomedical dataset

Classifier	ACC		EMG	
	MED	RMS	TD	TDPSD
LDA	99.9+0.3	96.3+5.2	63.0+9.7	62.3+8.0
QDA	99.9+0.1	98.4+1.8	67.8+8.9	66.0+7.6
*k*NN	**100.0+0.0**	98.0+2.4	66.8+8.1	54.8+8.4
RF	99.5+0.6	96.3+3.2	66.8+8.4	63.0+8.5

its predominant use in myoelectric control [13]. The EMG TD feature set was found to be best for the biomedical dataset whereas ACC MED was found to best for all HCI datasets. Further inspection of the performance of the EMG TD feature set with the LDA classifier is provided through the confusion matrices of the biomedical and HCI-A dataset in Table III. Conversely, Table III shows a similar confusion matrix using the best feature set determined for the HCI dataset (ACC MED). Finally, the results of sequential classification of gestures from multiple-positions are presented in Table IV.

DISCUSSION

This study corroborates the use of ACC as an accompanying modality in biomedical/clinical applications to achieve positional robustness. Table I verifies that accelerometers situated on the forearm and biceps can be used with confidence to decode 5 upper-limb positions in the sagittal plane. Despite encoding similar information from the ACC modality, the MED feature set consistently encoded positional information significantly better (p <0.05) than RMS. Table II provides an upper-limit of accuracy that can be achieved when position recognition is performed without fault. Use of a sequential classification framework achieved no statistical difference between the within-position gesture recognition framework when using ACC MED to segment position and EMG TD to recognize gestures. Although the position recognition performance of MED was statistically better than RMS, no statistical improvement is apparent in the gesture recognition accuracy of the sequential framework using these feature sets to decode position.

TABLE II: Within-position gesture recognition rates across positions

Dataset	Classifier	ACC		EMG	
		MED	RMS	TD	TDPSD
Bio	LDA	69.8+4.4	65.8+4.5	**96.2+0.7**	96.0+0.4
	QDA	66.4+4.8	64.3+5.1	95.1+0.8	94.2+0.5
	*k*NN	63.8+5.6	60.8+5.1	94.3+0.9	85.8+1.2
	RF	61.2+4.9	59.2+3.3	92.9+0.7	91.6+0.9
HCI-A	LDA	**97.1+1.5**	96.6+1.9	89.1+3.5	91.1+2.7
	QDA	93.8+3.8	89.0+5.5	82.9+5.3	68.4+7.0
	*k*NN	94.2+2.6	94.6+2.4	82.8+4.5	70.1+4.8
	RF	92.0+3.8	92.9+2.5	85.4+3.6	82.3+3.8
HCI-B	LDA	**94.4+4.0**	94.2+4.1	84.7+8.1	87.5+8.6
	QDA	88.5+8.5	84.4+8.5	75.0+8.4	53.1+10.4
	*k*NN	87.7+8.8	87.9+8.6	68.3+9.1	50.6+9.2
	RF	84.2+6.9	84.4+7.1	78.4+7.0	73.0+8.4
HCI-C	LDA	**89.1+4.4**	84.5+6.6	66.5+8.5	71.9+8.5
	QDA	87.9+8.1	84.1+8.9	60.9+9.6	45.9+8.8
	*k*NN	80.6+9.1	81.7+9.2	52.0+9.8	34.1+7.1
	RF	77.9+8.9	78.2+8.9	62.3+8.6	54.2+8.0

This study additionally corroborates the past outcomes of biomedical and HCI studies, where EMG is best for biomedical applications and ACC is best for HCI gesture recognition. Gesture recognition for biomedical applications, such as prosthesis control, relies on class-separability provided through EMG features (96.3%). Although ACC features provide moderate class-separability for the WS (83.9%) and WP classes (87.2%), they provide only marginal class-separability for other classes (mean: 55.7%). HCI gesture recognition results found that ACC features substantially outperformed EMG features with the same set of gestures (HCI-A), a set of finger gestures (HCI-B), and a set of grasping gestures (HCI-C). In contrast to past HCI experiments where 40 gestures are used together, the use of EMG TDPSD for gesture recognition with a subset of wrist gestures provided satisfactory accuracy (91.1%).

Although findings were consistent with past studies, there remains a disconnect between the use of ACC for the recognition of gestures between the biomedical and HCI frameworks. When no positional variance was purposely included (biomedical section of Table II), ACC provided no real gesture-specific information resulting in low accuracy. The high gesture recognition accuracy achieved using the HCI datasets is most likely an outcome of stratifying gestures across different positions to strategically reduce to improve the separability of the gestures. This use of positional variance can be seen in Fig. 1, where the HCI dataset shows distinct changes in ACC signals during contractions that are uncharacteristic of mechanomyography. This leveraging of positional variance was inferred in [3], where gestures performed "in the air" resulted in higher accuracy than gestures performed when in contact with a surface.

A limitation of this study is the use of static contractions alone in the biomedical dataset. Past studies have found that including ramp contractions can reduce the impact of contraction intensity variability by incorporating more dynamics [14]. It is possible that there may exist repeatable ACC patterns during the transient segment of such ramp contractions that could be leveraged as part of future multi-modal myoelectric control systems.

Ultimately, the consequence of different aims between biomedical and HCI applications can result in confusion when interpreting the outcomes of studies from both fields, especially the when terminology used to describe the gestures does not indicate the aim of the study. In light of this identified deficiency, it is suggested that a full review of past studies be conducted so as to develop a clear taxonomy and set of terminology that could be adopted by both of these expanding fields.

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TABLE III: Confusion matrix of the gesture classification accuracy (%) of the biomedical and HCI-A datasets using EMG TD and ACC MED feature sets using the LDA classifier

EMG TD												
True Label	Predicted Label - Biomedical						Predicted Label - HCI-A					
	WF	WE	WP	WS	PO	PI	WF	WE	WP	WS	PO	PI
WF	**97.4**	0.0	1.0	0.9	0.5	0.2	**94.4**	0.6	0.7	1.0	1.8	1.5
WE	0.2	**96.6**	0.1	0.7	1.8	0.5	0.2	**86.3**	7.4	1.7	1.6	2.8
WP	0.2	0.3	**96.2**	1.9	0.8	0.5	0.0	10.2	**83.5**	2.3	1.0	3.0
WS	0.1	0.0	1.0	**97.8**	0.7	0.4	0.3	1.0	2.6	**82.7**	10.9	2.4
PO	0.1	0.3	0.4	0.5	**95.2**	3.5	0.3	1.1	1.1	10.8	**85.0**	1.7
PI	0.3	0.1	0.5	0.5	4.0	**94.6**	0.1	0.6	0.5	0.4	0.1	**98.4**
ACC MED												
True Label	Predicted Label - Biomedical						Predicted Label - HCI-A					
	WF	WE	WP	WS	PO	PI	WF	WE	WP	WS	PO	PI
WF	**64.3**	15.8	2.1	3.2	8.8	5.7	**99.9**	0.1	0.0	0.0	0.0	0.0
WE	15.8	**59.6**	1.0	3.3	10.8	9.6	1.0	**97.5**	1.5	0.0	0.0	0.0
WP	4.0	2.8	**87.2**	1.0	3.9	1.1	0.4	6.4	**92.7**	0.5	0.0	0.1
WS	4.4	5.3	0.6	**83.9**	3.6	2.3	0.0	0.4	0.6	**95.1**	3.9	0.0
PO	10.0	9.8	2.7	2.7	**50.6**	24.3	0.0	0.0	0.0	4.8	**95.2**	0.0
PI	8.8	11.6	2.3	3.5	25.5	**48.2**	0.0	0.0	0.0	0.0	0.0	**100.0**

TABLE IV: Sequential gesture recognition accuracy (mean+std of subjects) of the biomedical dataset

Feature Set		Classifier			
Position	Gesture	LDA	QDA	*k*NN	RF
ACC MED	ACC MED	65.5+17.2	62.4+15.8	60.2+15.5	58.2+15.1
	ACC RMS	61.9+16.4	60.5+15.7	57.5+15.5	55.8+15.4
	EMG TD	**96.0+3.2**	94.7+4.3	93.9+4.2	92.5+4.0
	EMG TDPSD	95.6+3.5	93.6+4.0	84.7+5.5	91.0+4.6
ACC RMS	ACC MED	63.8+15.9	61.9+15.5	59.5+14.8	57.2+14.8
	ACC RMS	60.5+15.2	60.0+15.3	56.7+14.9	54.1+14.3
	EMG TD	95.5+3.1	94.4+4.2	93.6+4.3	91.5+4.2
	EMG TDPSD	95.2+3.4	93.3+3.8	84.3+5.4	90.6+4.5
EMG TD	ACC MED	50.2+12.0	49.7+11.1	51.1+13.0	47.3+11.5
	ACC RMS	46.7+11.0	47.4+11.0	46.6+12.5	44.1+11.6
	EMG TD	93.4+4.8	92.8+5.3	94.1+4.7	91.2+4.7
	EMG TDPSD	93.2+4.7	92.2+4.5	84.3+4.3	89.8+5.0
EMG TDPSD	ACC MED	49.7+11.7	49.1+11.2	47.0+10.8	46.5+11.5
	ACC RMS	45.8+10.5	47.3+11.1	42.0+9.9	43.4+10.8
	EMG TD	93.3+5.0	92.8+5.6	91.2+5.4	90.2+5.1
	EMG TDPSD	93.5+4.7	92.1+4.5	83.1+4.3	89.2+5.3

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A HOME-BASED MYOELECTRIC TRAINING SYSTEM FOR CHILDREN

Matthew Dyson [1], Jennifer Olsen [1], Kianoush Nazarpour [1, 2]

[1] *School of Engineering, Newcastle University, Newcastle-upon-Tyne, UK*

[2] *Biosciences Research Institute, Newcastle University, Newcastle-upon-Tyne, UK*

ABSTRACT

We present progress from an ongoing project which aims to develop a low cost home-use myoelectric training system for children. The training system is based on a first person game design within which children control a virtual limb and terminal device. Preliminary results indicate that the perceived level of control over the terminal device is high. However, designing a system which genuinely motivates and engages children remains a significant challenge.

INTRODUCTION

Children born with upper limb differences will typically reject a prosthesis unless it provides significant functional gain [1]. In the case of myoelectric prostheses a core factor which limits functional gain is control. The objective of this project is to develop a child-friendly game-based myoelectric muscle training system based on the principles of biofeedback. The system is designed for home-use and aims to be low cost. The assumption underlying our project is that myoelectric control can be implemented separately from a prosthetic device, allowing children to learn control before they are fit with a prosthesis.

It is widely recognised that patients usually fail to meet the number of movement repetitions required for behavioural change. Rehabilitation-relevant muscle activities in the context of game-play offer a motivational and engaging method to increase the amount of practise performed. Games can provide the challenging, intensive, task-specific conditions necessary to promote adaptation of behaviour [2]. In our training system players control a virtual limb with a simple terminal device. The objective in each level is to manipulate objects using a muscle decoding system based on [3]. The system does not attempt to simulate grasp but the avatar has anatomically correct dimensions and the game adheres to the principles of task-orientated gaming [4].

This work is part of an ongoing collaborative research project to co-design child prosthetics solutions [5]. As such, the work is not linear in nature. For the purpose of presentation, methods are split into two sections broadly outlining the first and second iterations of development.

METHODS

Ethics

All participants gave informed written consent. Approval was granted by the local ethics committee at Newcastle University (Ref: 17-NAZ-056).

Iteration One

Children played the game using two devices. The intact-limb controlled character movement in virtual space via a single-hand thumb stick. The virtual limb was controlled using a Shimmer3 EMG unit on the residual limb. Inertial measurement unit (IMU) data controlled the orientation of the virtual limb. Electromyography signals acquired from flexor carpi radialis (FCR) and extensor carpi radialis (ECR) controlled the virtual terminal device.

The game prototype uses a first person perspective. The core mechanics involve picking up and manipulating objects in a scene. Participants progressed through six levels. The first and second tutorial levels introduced EMG, IMU and combined EMG and IMU control. The three main levels of the game were themed around teaching a) delicate object manipulation, b) directed muscle co-contraction and c) extended manipulation of objects to reach a goal. A final optional level introduced a competing non-player character to limit the time available to complete tasks.

The system was tested on four children, two of whom had trans-radial limb deficiencies. After playing, children, and optionally their parents or guardians, answered a short questionnaire about perceived control and provided open feedback on the game in general. Feedback was also solicited from relevant domain experts.

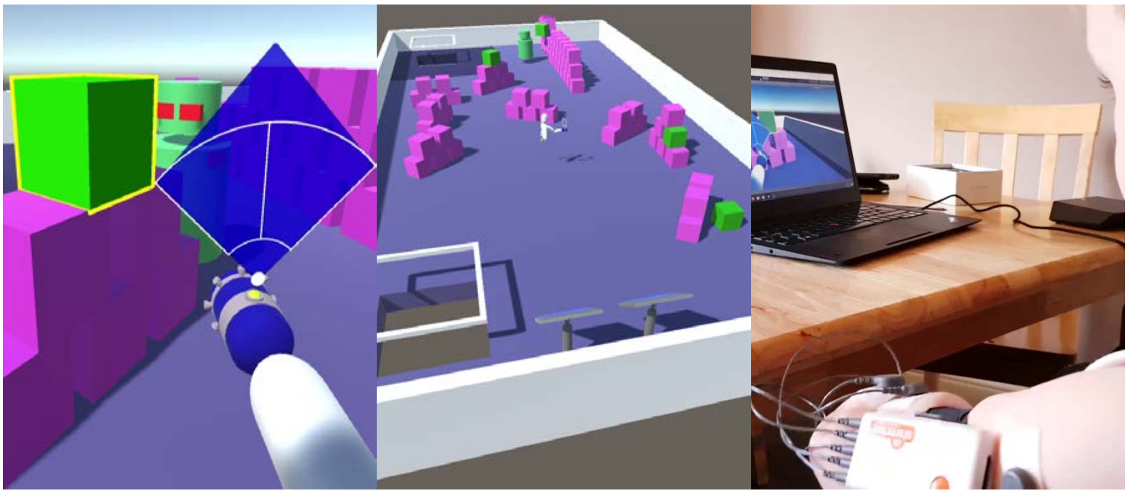

Figure 1: Iteration one. Left: play perspective showing virtual limb and biofeedback panel. Middle: a game scene, player (white) competes with non-player character (green) to collect blocks in the environment. Right: participant playing the game in a home environment.

Iteration Two

The first iteration of the game environment was built using game engine primitives. The graphics in the second version use purchased assets to provide a modern visual aesthetic, shown in Figure 2. Based on expert feedback, the game scoring systems and general time limits were updated to create a greater sense of challenge in the tasks.

The second iteration of the game uses a Delsys Quattro sensor for experimental data acquisition and trials a custom microprocessor-based controller for longer-term testing, shown in Figure 2. The microprocessor controller uses dry electrodes and performs all the signal processing necessary to send control signals to the virtual game limb. All personalised settings are retained on the device and shared with the game on connection. As EMG processing only involves linear filters, the controller side-steps sampling issues associated with low-cost EMG systems [6].

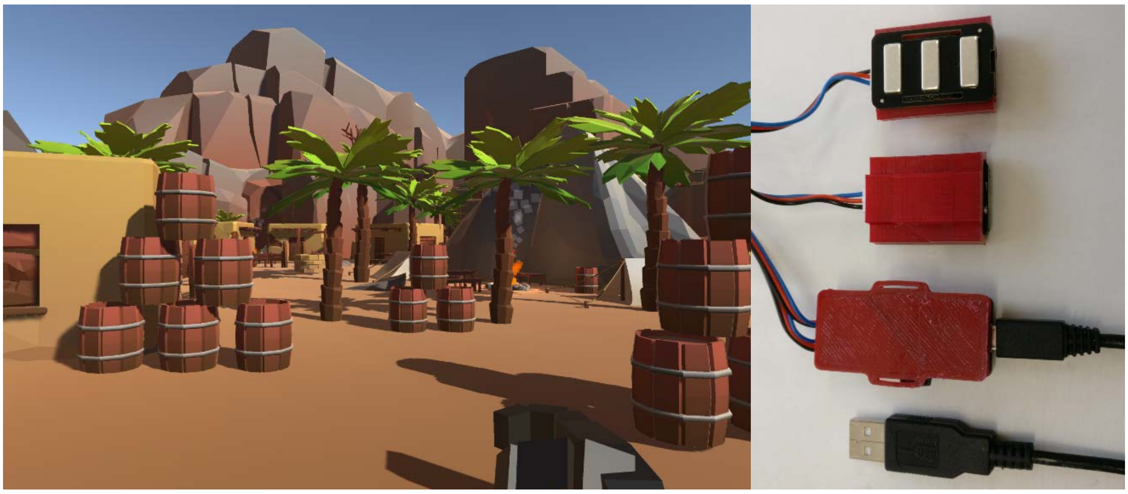

Figure 2: Images from iteration two. Left: game scene based on purchased assets. Right: microprocessor game controller, signal processing and IMU unit and two dry EMG electrodes.

RESULTS

Iteration One

Results of the perceived control questionnaire are shown in Table 1. The general rating of control for the player avatar, the virtual limb and the virtual terminal device were positive.

Table 1: Children's rating of control of game environment character.

Participant		***Rating of Control (1 poor to 5 good)***		
#	***Amputee***	***Robot***	***Arm***	***EMG***
1	Y	5	5	4
2	Y	5	5	4
3	N	4	4	3
4	N	5	4	5
Average		**4.75**	**4.5**	**4**

In general, the open feedback focussed on proposals for creating more engaging game experiences, with the majority of children providing relatively specific recommendations which would make the game better for themselves. When probed three out of four children indicated they found the tasks in each level too tedious to consider performing repetitively.

The use of gel-based snap electrodes caused a number of issues during data collection for iteration one, particularly when working with younger children. In most cases the time required to place electrodes, especially on smaller limbs caused frustration and often raised questions from parents / guardians about real world practicality.

Iteration Two

Preliminary tests for iteration two have focussed on comparisons of control systems. Development has aimed to obtain a degree of parity between the microprocessor controller and the Shimmer EMG device.

The two controllers use different EMG sensors, acquisition rates, and desktop PC interfaces, therefore comparisons have been made solely on perceived user preference. Tests were run an ad-hoc and informal basis using EMG naïve able-bodied adult participants. During tests, participants were aware of the context of the research. Participants moved virtual blocks using the game platform developed as part of iteration one and were asked for feedback as to which device they preferred.

Table 2: EMG and IMU preferences when comparing microprocessor and Shimmer controllers.

Participant	***Blocks Moved***		***Control Preference***	
#	***Microcontroller***	***Shimmer***	***EMG***	***IMU***
1	8	6	No preference	Shimmer
2	6	8	No preference	No preference
3	5	7	No preference	Shimmer

Result of recent tests are shown in Table 2. No participants expressed an EMG control preference. Two of three participants expressed a clear preference for using the Shimmer device to control the position of the virtual limb.

DISCUSSION

While children rated their overall level of control as high, perception of overall potential for engagement was subjectively low. This problem is not unique to game-based systems orientated toward children, it likely reflects a more general issue inherent to attempting to designing video games for rehabilitation purposes [2]. While the barrier of entry to creating games is now low, the skills necessary to design engaging games remain within a small group of dedicated professionals catering for larger markets. In the context of game-based rehabilitation for children, these problems of motivation and engagement are further compounded by the challenge of ensuring any behavioural activities involved are appropriately task orientated [4, 7].

Recent research questions the assumptions which typically underpin game-based training systems for prosthetics, instead proposing that transfer of learning from a virtual task to real world use only occurs when the coupling of action and perception is matched between tasks [7]. In the context of learning myocontrol, this places greater importance in replicating end-effector behaviour when reaching for, grasping and manipulating objects. How best to provide this feedback at a low cost and with relatively low complexity for younger children is unknown. Current age recommendations for virtual reality devices err on the side of caution, as such the most appropriate platform for simulating perception in adults will not necessarily be available for children in the near future.

When considering paediatric upper-limb prosthesis rejection rates [1] and the effective age ranges for rehabilitative intervention [8] it appears highly unlikely that any one game-based rehabilitation technology would be suitable for all children. A more productive approach may therefore focus on enabling the necessary hardware platforms to deliver effective child-appropriate game-based rehabilitation. As the overall market size for this type of technology remains limited, it may be prudent to consider designing for other appropriate paediatric use cases.

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INEXPENSIVE SURFACE ELECTROMYOGRAPHY SLEEVE WITH CONSISTENT ELECTRODE PLACEMENT ENABLES DEXTEROUS AND STABLE PROSTHETIC CONTROL THROUGH DEEP LEARNING

Jacob A. George, Anna Neibling, Michael D. Paskett, Gregory A. Clark

Department of Biomedical Engineering, University of Utah, Salt Lake City, Utah, USA

ABSTRACT

The dexterity of conventional myoelectric prostheses is limited in part by the small datasets used to train the control algorithms. Variations in surface electrode positioning make it difficult to collect consistent data and to estimate motor intent reliably over time. To address these challenges, we developed an inexpensive, easy-to-don sleeve that can record robust and repeatable surface electromyography from 32 embedded monopolar electrodes. Embedded grommets are used to consistently align the sleeve with natural skin markings (e.g., moles, freckles, scars). The sleeve can be manufactured in a few hours for less than $60. Data from seven intact participants show the sleeve provides a signal-to-noise ratio of 14, a don-time under 11 seconds, and sub-centimeter precision for electrode placement. Furthermore, in a case study with one intact participant, we use the sleeve to demonstrate that neural networks can provide simultaneous and proportional control of six degrees of freedom, even 263 days after initial algorithm training. We also highlight that consistent recordings, accumulated over time to establish a large dataset, significantly improve dexterity. These results suggest that deep learning with a 74-layer neural network can substantially improve the dexterity and stability of myoelectric prosthetic control, and that deep-learning techniques can be readily instantiated and further validated through inexpensive sleeves/sockets with consistent recording locations.

INTRODUCTION

Neural networks have been used to classify hand gestures from surface electromyography (sEMG) with high accuracy [1]. However, these improvements have not been realized for kinematic regression, where the network is used to control multiple degrees of freedom (DOFs) simultaneously and proportionally. Although neural networks are effective at suppressing unintended movement (i.e., reducing cross-talk), their proportional control is noisy, which ultimately can make neural networks inferior to Kalman filters in functional tasks [2]. One explanation for this poor performance is that the amount of data used to train neural networks in past work was roughly two orders of magnitude less than what is traditionally used for deep learning in other domains, and performance is critically dependent on large training datasets [3].

Gathering large datasets of sEMG synchronized to motor intent is particularly challenging because patient time is limited and the placement of recording electrodes changes day to day [4]. Here, we demonstrate a simple approach to gather large datasets of sEMG and thus enable deep learning for myoelectric prostheses. We first introduce an inexpensive sEMG sleeve that can be repeatedly donned with consistent electrode placement, and then we demonstrate how accumulating spatially consistent sEMG over time yields the large datasets necessary for deep learning. The results of this case study suggest that deep learning can improve the dexterity and robustness of myoelectric prostheses.

METHODS

Sleeve Design and Fabrication

The sEMG sleeve was constructed from neoprene fabric sewn into a hollow cylindrical shape after electrodes and wires were inserted (Fig. 1A). The neoprene can be stretched during donning and doffing, but also provides enough structural integrity to maintain consistent placement on the forearm. Brass-coated marine snaps served as inexpensive dry electrodes; 32 were embedded across the full circumference and length of the sleeve to record from extrinsic flexors and extensors. Two additional electrodes were embedded at the proximal end of the sleeve to be placed along the ulna bone and serve as an electrical reference and ground. Each electrode was soldered to a segment of flexible wire with high-strength heat shrink to reduce wire breakage. Electrodes were embedded into the neoprene using a crimping tool, and loops of wire were formed to near each electrode to alleviate strain when the fabric is stretched. Wires were stitched onto the neoprene and soldered to a 38-pin SAMTEC connector. Grommets were inserted into the neoprene, unique to one intact individual, such that the grommets aligned with natural skin markings (e.g., freckles, moles, scars) (Fig. 1B). A loose cover made of Lycra® was used to electrically isolate wire and house front-end devices for amplification and filtering (Fig. 1C). The sleeve can be manufactured in a few hours and costs less than $60 (Table 1).

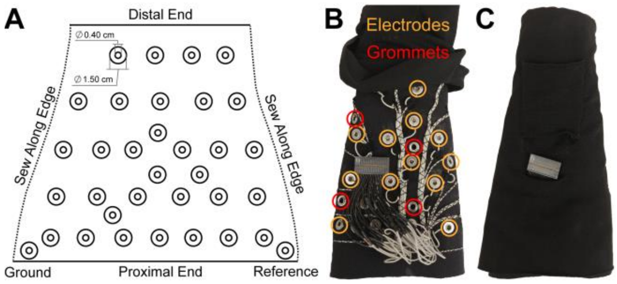

Figure 1: Inexpensive surface electromyography sleeve with consistent electrode placement. A) Thirty-four electrodes (brass-coated marine snaps) are inserted into neoprene, and then the fabric is then sewn into a sleeve. B) Grommets (red) are inserted alongside electrodes (orange) so they align with natural skin markings (e.g., freckles, moles, scars). C) A loose Lycra® cover electrically isolates wire.

Table 1: Cost of Materials

Component	Description / Use	Cost
Coated brass marine snap fasteners (34)	Dry recording electrode to record surface electromyography	$8.83
Round stainless-steel washer (34)	Electrode backing to hold electrodes into sleeve	$3.07
Neoprene fabric (1.5 sqft)	Stretch material for easy don/doff	$2.98
Lycra fabric (1.5 sqft)	Cover material to reduce electrical noise with movement/contact	$1.60
Copper wire (26 ft)	Electrode connections	$17.16
Heat shrink (1.5 ft)	Reinforcement for solder joints	$1.62
Thread (3 ft)	Assembly of fabric and wiring	$0.01
SAMTEC connector (1)	Connection to front-end amplification and filtering	$23.20
Total:		**$58.47**

Signal Acquisition

Thirty-two monopolar sEMG electrodes were sampled at 1 kHz using Micro2+Stim Front-Ends and a Grapevine Neural Interface Processor (Ripple Neuro LLC). The 300-ms smoothed Mean Absolute Value (MAV) on the 32 single-ended electrodes (or 528 possible differential pairs) was calculated at 30 Hz [5]. Signal-to-noise ratio (SNR) was defined as the mean 300-ms smoothed MAV during movements (see Training Datasets below) divided by the mean 300-ms smoothed MAV during rest.

Sleeve Performance

Seven intact participants were recruited to validate the sleeve performance. All experiments were performed with informed consent and under protocols approved by the University of Utah Institutional Review Board and the Department of Navy Human Resources Protection Program. Each participant donned the sleeve five times, attempting to align the grommets with colored markings on their forearms. The times to don and doff the sleeve, as well as the average distance between donned positions, were recorded. Participants also completed one training dataset to determine the sEMG SNR.

Training Datasets

sEMG and intended movement were recorded simultaneously while participants mimicked the following six preprogrammed movements of a virtual prosthetic hand (MSMS [6]): flexion/extension and abduction/adduction of D1; flexion/extension of (D2); simultaneous flexion/extension D3, D4 and D5; and flexion/extension and pronation/supination of the wrist [5], [7]. Seven intact participants completed one training dataset to determine the sEMG SNR. For one intact participant, a total of 20 datasets were collected over time, each requiring the sleeve to be donned and doffed. For this participant, each dataset introduced slight variations in the movement speed, movement hold-time [2], and forearm posture.

Neural Network Control Algorithms

Two neural networks were used in this study. The first was a shallow, 10-layer, neural network with similar input and architecture as [2], but with added 50% dropout layers after each rectified linear unit. The second network was a deep, 74-layer, residual neural network with similar architecture as [3]. Input at each timepoint consisted of the MAV from 32 single-ended electrode recordings over the last 32 time-samples (~1.07 seconds) (Fig. 3A). The bulk of the network consists of nine residually connected convolutional units, each of which consisted of two repetitions of a 3x3 convolutional layer followed by batch normalization, followed by a rectified linear unit (Fig. 3B). The output of both neural networks was the kinematic predictions for the six-DOF virtual prosthetic hand. An optional unmodified Kalman filter [5] was placed on the end of deep neural network to smooth the kinematic predictions.

Because tests were performed entirely online, the networks were trained with 97% of the training data, and the remaining 3% was used for validation to avoid overfitting. Training automatically terminated once the root-mean-squared-error (RMSE) on the validation data increased to avoid overfitting. The networks were trained using a Stochastic Gradient Descent with Momentum solver with an initial learning rate of 0.001 [2].

Online Performance Metrics

The participant completed a real-time virtual hand matching task in which they actively controlled the virtual prosthetic hand and attempted to move only select DOF(s) to a target location [5]. Performance was evaluated as the mean longest continuous-hold duration (i.e., hold duration) within the 10%-error window around the target location out of a theoretical maximum of seven seconds (i.e., seven seconds max if no reaction time) [5]. Performance of the shallow network, trained only on the first dataset, was evaluated ten times over the span of 263 days. Performance of the shallow network trained on the first ten datasets was also directly compared against the performance of the shallow network trained on only a single dataset collected immediately before the task. Prior results demonstrated that Kalman filters produce smoother movements than neural networks, which is critical for functional tasks [2]. To this end, we evaluated the performance of a deep neural network trained on 20 datasets with and without a Kalman filter to smooth the network output. Direct comparisons were performed using a counterbalanced pseudorandom cross-over design.

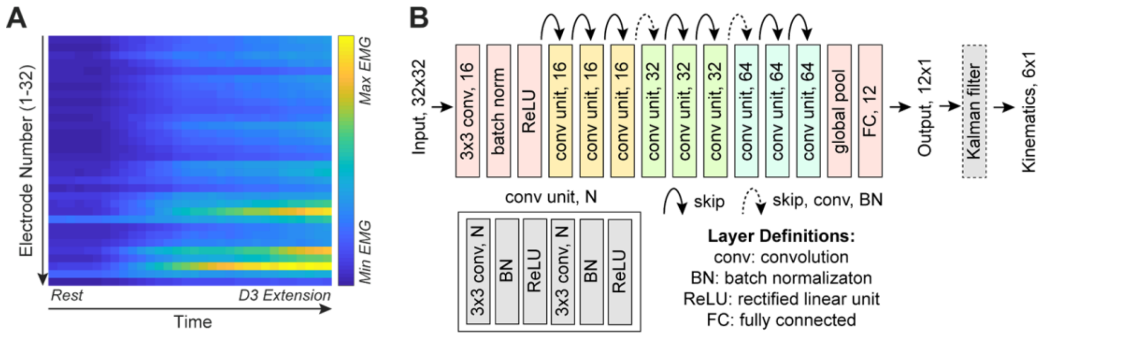

Figure 2: Deep residual neural network for myoelectric prosthetic control. **A)** Example 32-by-32 input "image" consists the 300-ms smoothed MAV on the 32 single-ended electrodes over the last 32 time-samples (i.e., the last ~1.07 seconds). The example image shows EMG activity increasing across all channels as the participant transitions from rest to D3 extension, although EMG activity is highest on channels 23, 28, and 30. **B)** Multiple layers of convolution across electrodes and across time allows for complex non-linear activation patterns. An optional Kalman filter was used to smooth the kinematic predictions from the deep neural network.

RESULTS

Inexpensive sleeve enables consistent placement and prosthetic control

Seven intact participants were able to self-don the sEMG sleeve within 7.32 ± 0.26 mm of precision in 10.30 ± 3.35 s (Fig. 3). The mean SNR for participants was 14.03 ± 4.43 (Fig. 3). A shallow neural network, trained on only a single dataset one day before the start of testing, provided relatively stable performance over time in one individual tested longitudinally (Fig. 4A). Performance fluctuated over time ($p < 0.05$, one-way ANOVA), but the performance on day one was not significantly different from that on any subsequent day, including 263 days after initial training (p's > 0.05, multiple pair-wise comparisons with correction).

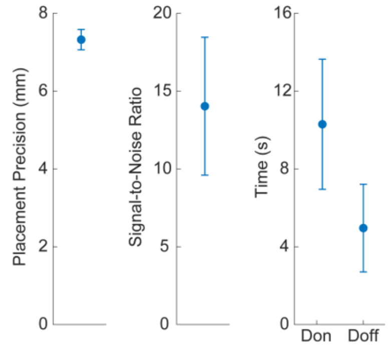

Figure 3: The inexpensive sleeve can be donned rapidly with sub-centimeter precision and adequate signal-to-noise. Data show mean ± S.E.M. for seven intact participants.

Deep learning substantially improves prosthetic control

We hypothesized that additional training data would improve neural network performance. To this end, we compared the participant's performance with a shallow neural network trained on a single dataset collected immediately before testing against a shallow neural network trained on 10 datasets collected multiple weeks prior. Performance (hold-time duration) doubled when trained on 10 prior datasets ($p < 0.05$, paired t-test; Fig. 4B).

Deeper neural networks have a greater capacity to learn the intricacies of large complex datasets. Building on this idea, we trained a deep neural network on 20 prior datasets (Fig. 2). Performance significantly improved relative to the shallow networks reported here (p's < 0.05, paired t-tests). Deep neural network performance further improved when a Kalman filter was added to the end to smooth kinematic predictions ($p < 0.05$, paired t-test; Fig. 4C). The final architecture, a deep residual neural network with a Kalman filter (DNN+KF), resulted in up to a 152% improvement relative to a modified Kalman filter (4.42 s reported here vs 1.75 s reported previously with the same task and the same participant [2]). Informally, the participant used the DNN+KF to control a physical prosthesis (LUKE Arm) and was able to grasp objects while simultaneously rotating and flexing the wrist – a task that is particularly challenging when simultaneously controlling the position of six different DOFs.

CONCLUSION

This work first highlights an inexpensive sEMG sleeve with consistent electrode placement. Then, we show how these consistent recordings can enable deep learning, and drastically improve dexterity and stability of myoelectric prosthetic control. Although this latter finding is based on the results from a single participant, the findings are consistent with a recent study that also used spatially consistent sEMG accumulated over time to improve neural networks [8]. Future work should expand this approach to a larger cohort of amputee participants, and validate dexterity and stability in activities of daily living. Additional training paradigms or design considerations may be necessary to account for sEMG variations due to sweat, swelling, and/or excessive fatigue.

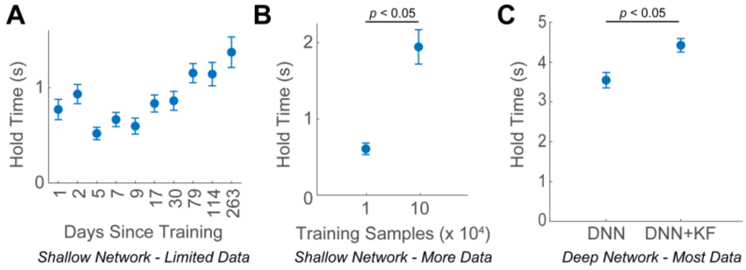

Figure 4: Online performance of prosthetic control. Neural networks were trained to predict motor intent based on surface electromyographic recordings from a custom-made sleeve that maintained consistent electrode placement. **A)** Functional performance, evaluated as the ability to hold complex grasps for up to seven seconds [2, 5], was relatively stable over time. Performance on day one was not significantly different from that on any subsequent day, including 263 days after initial training (p's > 0.05). **B)** A neural network trained on 10 prior datasets accumulated across time doubled the performance of a neural network trained on single dataset from the current day. **C)** A deep neural network (DNN) trained on 20 prior datasets further improved performance relative to the shallow networks in B (p < 0.05). Adding a Kalman filter to smooth the output of the DNN (DNN+KF) significantly improved performance yet again. DNN+KF performance was 152% greater than what was previously reported for a modified Kalman filter [5] with the same participant [2]. Data show means ± S.E.M. from one participant. p-values shown for paired comparisons.

ACKNOWLEDGEMENTS

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INTUITIVE OPTIMAL PROSTHESIS TUNING THROUGH USER TUNED COSTS OF EFFORT AND ACCURACY

Anjana Gayathri Arunachalam, Kevin B. Englehart, and Jon W. Sensinger
Institute of Biomedical Engineering, University of New Brunswick

ABSTRACT

Clinicians and prosthesis users care about the practical attributes of movement like the physical effort required by the user, the response time of the device, the reliability, and the accuracy of the movement performed. But the calibration parameters of a prosthetic device are relatively abstract and do not directly correspond to those quantities that the users and the clinicians inherently care about. Here, we propose an intuitive tuning technique that allows clinicians to tune prostheses based on the things that end-users actually care about. We use well-established engineering techniques (optimal control) to determine the set of best possible solutions for different relative preferences of the user. This required optimizing the problem for multiple objectives, (effort, time, reliability, and accuracy) to compute the best tuning parameters for a wide range of trade-offs. By solving this optimization problem, the complexity of the relationship between the performance and the prosthesis parameters can be implemented as a mapping procedure, and thereby hidden from the user. This simplifies the calibration process and allows clinicians or users to intuitively customize the device for their individual needs.

INTRODUCTION

Biological movement and motor coordination can be thought of as optimization tasks that minimize the cost of effort and time while maximizing the reward obtained from performing the movement [1], [2]. The costs are mathematical representations of quantities that the brain tries to minimize when generating any motor command to move our body. Several different cost functions like effort, metabolic energy, and endpoint variance have been used to describe specific movements. This inconsistency in literature actually suggests that humans optimize a combination of different costs [3] and simply change their cost priorities to perform different tasks. The composite costs of effort, accuracy, reliability and time sufficiently describe how we consistently coordinate our joints to perform different tasks [4]. This cost preference also changes from person to person. Suppose we ask a group of people to write by hand the entire abstract of an article- some might care more about the time spent on the task and write as fast as they could, while others might care more about the reliability of the outcome and don't mind spending a few extra minutes.

Clinical motivation

Let's take the example of driving a car. We usually care about things like fuel efficiency, comfort, safety, and the dynamic response of the car. The input parameters like the steering force, powertrain characteristics, suspension control, two-wheel and four-wheel drive modes can be adjusted to reflect our personal priorities in terms of the things we care about when driving. An experienced driver might be able to easily tune these parameters to get the desired response. But for new drivers, this could be a daunting task. The driving mode options provided by most car manufacturers nowadays, simplifies this task for both new and experienced drivers. The different modes like the eco mode, comfort mode or the sport mode speak the language of the user and directly convey the information in terms of things they care about.

Likewise, both clinicians and end-users of prosthesis care about the costs of effort, time, accuracy, and reliability incurred by the users when making a movement with a prosthetic device. The input parameters for a myoelectric prosthesis are abstract quantities like the device gain, amplifier thresholds, and control mapping paradigms like proportional position or velocity control. But unlike the example with the cars, the clinicians are not provided with a set of "driving mode" settings that simplifies the relationship between the cost space in which they care about and the input parameter space in which they work.

Moreover, there is an inherent trade-off in the balance of these cost preferences. That is, we cannot have the best of both worlds and improve both the speed and the accuracy of our movements simultaneously. When larger control signals are produced to make faster movements, the multiplicative nature of the noise in our myoelectric signals deteriorates the accuracy and the reliability of the movement. The device gain parameter should hence be adjusted to a sweet spot that best reflects the user preference. But there is another catch, some combinations of the input parameters always produce results that are worse for all possible user preferences. For example, proportional velocity control always performs better than position control in terms of both the costs of effort and reliability. Conventional tuning techniques force the clinicians to tune a limited number of abstract parameters that do not directly reflect the cost space that they care about, and make it much harder to tune for the optimal set of parameters for a given user.

So when tuning a prosthetic device, it will be beneficial to avoid the sub-optimal input parameters to ensure that the user gets the best experience. Optimization methods or optimal control techniques can be used to identify the optimal input parameters for each individual user preference. This approach makes the tuning procedure much more intuitive for the clinicians as the abstract device input parameters can be computed and set by an algorithm that optimizes based on the costs that users inherently care about.

Background

We care about a variety of things when making a movement. As we have multiple objectives that we wish to optimize, we need to perform multi-objective optimization (MOO) to find the best set of tuning parameters for each user's personal preference. For a MOO, no single solution can be best with respect to all the conflicting objectives and we have several optimal solutions instead. The optimal solutions are those in which we cannot further reduce the cost of one objective function without increasing the cost of another. These optimal solutions are called the Pareto solutions, or the Pareto set [5].

There are two broad strategies for obtaining these Pareto solutions for MOO problems.

1) The first method is to scalarize the different objectives and to repeatedly solve for the entire range of cost- preferences.

   In terms of our prosthesis tuning example, this would mean that the different objective functions of effort, time, reliability, and accuracy get added up with relative weights that represent the user's preference for the different cost functions. The computation is then repeated for every user's individual preference.

   For example, if the total cost (J) is represented as a sum of two independent costs (J_1 and J_2):

   (i) $J = J_1 + J_2$, represents an equal preference for the two costs.

   (ii) $J = J_1 + 10\ J_2$, represents that the user cares about 10 times as much about the second cost when compared to the first one.

2) The second method is to find multiple Pareto-optimal solutions in a single run, without any prior information about the relative preference of the different costs.

   For our prosthesis tuning problem, this would mean that we compute a set of optimal tuning parameters without any reference to the user's individual preference for the different costs. This method can be represented mathematically as:

   J = [J1, J2], where we simultaneously try to optimize for both the cost functions J1 and J2.

The first strategy expresses the user's preference in terms of simple relative weights that change the optimum tuning parameters for the individual. But the tuning parameters must be recomputed every time the user's preference changes. The interpretation of relative weights also becomes incorrect when the multiple objective functions are not normalized appropriately [6]. The second strategy has the advantage of solving multiple optimal tuning conditions in a single simulation run, but it is not possible to incorporate the user preference into the algorithm. Our third option is optimal control which best approximates how humans move their joints and control human-machine interfaces [7], but the disadvantage is that this method requires a single objective function. Hence, we decided to blend the three ideas to get multiple meaningful Pareto solutions for the tuning parameters such that they can be saved in a look-up table to avoid re-evaluation.

Technical challenge

There are infinite solutions that satisfy the Pareto optimality condition and form the Pareto front. Ideally, we would like to obtain the optimal tuning parameters for a finite number of points on the Pareto front. These Pareto solutions can be saved in a look-up table that can aid with prosthesis tuning. Due to the inherently nonlinear nature of the human movement cost functions, an equally spaced set of relative weights does not produce a uniform Pareto set. Figure 1 shows an example in which only the costs of effort and accuracy of the movement were considered, for a first-order dynamic model of the prosthetic device. In this case, the Pareto solutions found are clustered towards one end of the Pareto front in which the effort cost is much lower than that of the accuracy cost. This means that a step-change in the user's preference for the two costs will lead to a larger change in the cost of effort when compared to the cost of accuracy. This also indicates that the problem is not accurately normalized (and that it requires nonlinear normalization mapping).

To describe the priorities meaningfully in terms of the cost, we need to flip the problem on its head and obtain a set of evenly distributed Pareto points that correspond to specific user priorities. This will allow us to bin the different regions of the Pareto front into “tuning modes” similar to the driving mode options provided by car manufacturers.

In addition to conveying the tuning information in the language of the user, this technique allows us to entirely avoid normalization. In this article, we propose a gradient-based approximation technique that can be used to produce an evenly distributed set of points on the Pareto front.

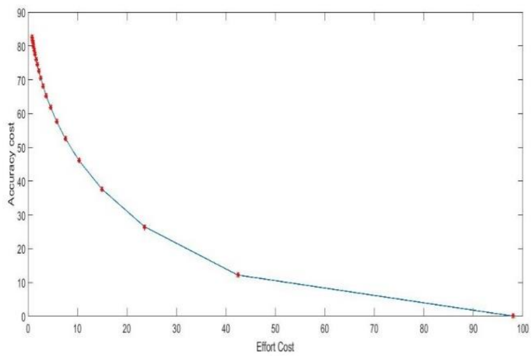

Figure 1: Pareto front for a set of cost preferences linearly spaced between 0.01 and 10. The red asterisks indicate the location of the obtained Pareto set.

METHODS

The main aim of this study is to determine if a uniform set of Pareto solutions can be obtained for a generic optimal control problem that has multiple objective functions. Optimal control guarantees the best solution, but it requires a single objective function, which is at odds with our attempt to enable users engage with multiple objectives. To solve this problem, we strategically assign weights across the multiple objectives to enable them to be considered as a single objective (which can then be solved using optimal control). Because this scalar composite cost can be solved using optimal control, it is guaranteed to land on the Pareto front, but where it lands on the Pareto front depends on the weights we choose. In order to strategically assign those weights to ensure an evenly distributed set of Pareto front, we use a crowding metric to decide where on the Pareto front we would like to land next, and then estimate the weights that should get us in that ballpark using a gradient-based approximation technique. This process is further described in the section on the Gradient-approximation technique.

To demonstrate the feasibility of the blended optimal control and MOO based approach, we use a simple optimal control example that relates to our ultimate aim of clinical prosthesis tuning. The example is that of a simple human-machine interface that is used to perform a target reaching task. In order to keep the problem simple and retain the multi-objective nature of human movement, only the two contrasting objectives of effort and accuracy were used to compute the cost incurred to the user. The human-machine interface model contained two parts: the human component that modelled the user’s motor commands, and the prosthesis component that performed a reaching movement in response to the user’s control signal. The system was assumed to be deterministic and the potential uncertainties in the control signal and the environment were not modelled.

The human component of the model produces an optimal control signal (u) that minimizes the composite cost of effort and accuracy to the user. The machine or the prosthesis component was simulated using standard zero, first or second-order dynamic system models. The tuning parameters of the prosthesis were not optimized in this study to reduce the number of optimization parameters. The duration of the movement was fixed to be a single time step for all conditions. The model was entirely implemented in MATLAB (Release 2016a, The Mathworks, Inc., Natick, MA.)

Cost Function and User Priority

The cost of effort was defined as the squared control signal u that represents the magnitude of the myoelectric signal from the user.

$$J_u = u^2 \qquad (1)$$

The cost of accuracy penalizes based on the error between the target (represented as g) and the movement endpoint.

$$J_a = [g - x(p)]^2, \qquad (2)$$

where $x(p)$ is the position of the simulated device at the end of the movement and the final time p is set to one for all the simulations without loss of generality.

The total cost is a weighted sum of the accuracy and the effort costs and is represented by:

$$J = \alpha . J_u + J_a, \qquad (3)$$

where α shows the user’s relative preference for the two costs. A large value of α shows that the user would rather minimize their physical effort even if it means that they don’t reach the target accurately. A small α value indicates that the user prioritizes the endpoint accuracy and won’t mind spending more effort. For a perfectly normalized set of costs, an α

value of one will indicate that the user cares about the two costs equally. But as the magnitudes of the cost values are highly task-dependent, normalization was not performed for our system.

Gradient-based approximation technique

The purpose of this algorithm boils down to two simple things – selecting the next point on the Pareto front that needs to be populated and landing there by computing the required α value. The distance (Mahalanobis form) between consecutive points was used as a measure of crowding and the next point was selected to ensure an even distribution of points on the Pareto front. We need to compute the desired user preference level or the α value to land at these points and a simple gradient approximation technique was used to achieve this. The required user preference value was calculated using the following equation.

$$\propto_{Desired}=\frac{\left(J^{i}_{Desired}-J^{i}_{Current}\right)}{\frac{\partial J^{i}}{\partial \alpha}\big|_{J^{i}_{Current}}}+\propto_{Current}\quad , \qquad (4)$$

where J^i corresponds to the individual costs of the effort or accuracy objective functions, and $\frac{\partial J^i}{\partial \alpha}\big|_{J^i_{Current}}$ refers to the sensitivity of the individual cost with respect to changes in α, computed at the current Pareto solution. These sensitivities are obtained by perturbing the α values at the different Pareto solutions obtained and are essentially just numerical approximations of the gradient at those points.

The two extreme points on the Pareto front that correspond to slopes 0.01 and 100 were picked heuristically by tuning the α values for the given task. After this, the “selection” and the “landing” algorithms were used iteratively to obtain a uniformly distributed Pareto set.

RESULTS

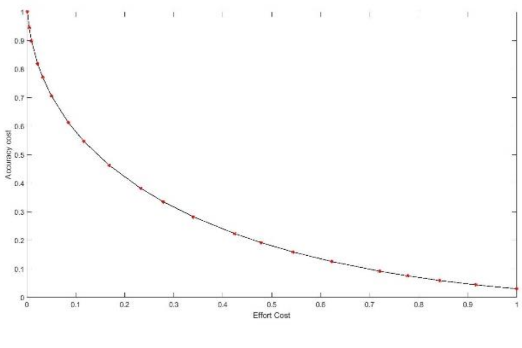

Figure 2: Uniform Pareto front obtained using the proposed algorithm

In order to ensure that a uniform Pareto front can be produced using our algorithm, a variety of scenarios were tested. The prosthesis was modelled as a standard zero, first, or second-order dynamic system. The proposed algorithm was able to successfully produce a uniform distribution on the Pareto front for all three cases. Figure 2 shows an example of the Pareto solutions found for a first-order dynamic model. These results show that the algorithm is generalizable and can be applied to optimize the tuning parameters for a variety of different user preferences.

DISCUSSION

The aim of this study was to determine if we can generate sufficient Pareto solutions for a human-machine interaction model, such that we can adequately describe different user preferences. Our simulations demonstrate the feasibility of the proposed method and show that it is robust for a variety of scenarios. The concept of relative tuning that we have described in this article could allow intuitive prosthesis calibration in terms of quantities that the clinicians and the patients care about. It will also permit the device to be tuned by the user. The user-tunability function can allow them to optimally perform vastly different tasks like painting and yard work, which requires them to change their personal cost priorities.

Limitations

As the intention of this article was to understand if a uniform set of Pareto solutions can be formed, the mathematical model was simplified to reduce the computational complexity of the problem. For example, the system was assumed to have no noise and the tuning parameters of the prosthesis model were not optimized.

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REAL-TIME PATTERN RECOGNITION OF FINGER MOVEMENTS USING REGENERATIVE PERIPHERAL NERVE INTERFACES AND IMPLANTED ELECTRODES

Alex Vaskov[1], Philip Vu[2,4], Alicia Davis[3], Theodore Kung[4], Paul Cederna[4], Cynthia Chestek[1,2]

1. Robotics Institute, University of Michigan

2. Department of Biomedical Engineering, University of Michigan

3. Orthotics and Prosthetics Center, University of Michigan

4. Department of Surgery, Section of Plastic Surgery, University of Michigan

ABSTRACT

Commercial myoelectric control systems using surface electromyography are unable to obtain consistent control signals for finger-specific motions because the desired signals are either obscured by more superficial muscles or non-existent due to the level of amputation. Intramuscular recording techniques and Regenerative Peripheral Nerve Interfaces (RPNIs) can potentially resolve each of these issues. Two persons with transradial amputations had bipolar electrodes surgically implanted into residual musculature and RPNIs. Participants used a low latency pattern recognition system to intuitively distinguish 7 individual finger postures with 100% online success and complete a functional task requiring multiple grasps with a commercially available prosthetic hand. A classifier with the same architecture was also used to distinguish movements in a simultaneous and proportional 2 degree of freedom control scheme. Both participants used this controller in real-time to complete a virtual target matching task with success rates of 99%.

INTRODUCTION

Traditional myoelectric prostheses for persons with upper-limb amputations are controlled by residual muscle activity via electromyography (EMG) recorded from the skin surface. Pattern recognition systems seek to provide users with intuitive control of wrist and hand functions. However, grip selection remains unintuitive as control is limited to simple open/close due to the lack of robust signals specific to finger movements [1]. Surgical interventions such as Targeted Muscle Reinnervation can create additional motor control sites [2] and more recent research has demonstrated the potential to extract specific motor inputs with signal decomposition [3]. Focusing on movement transitions has also allowed researchers to demonstrate more intuitive switching between a few grips [4]. However, without direct access to muscles that control fingers these techniques rely on algorithms to distinguish individual finger movements from subtle co-activations of prominent muscles or highly obscured deep muscle activity. Therefore, more work is needed to demonstrate that these techniques generalize outside of controlled tests. Given these challenges, it is also not surprising that pattern recognition is very sensitive to surface electrode placement [5]. Instead of attempting to resolve these issues with software alone, this study evaluates the use of intramuscular electrodes which can record large amplitude movement-specific EMG when implanted directly into finger flexors and Regenerative Peripheral Nerve Interfaces (RPNIs).

RPNIs are created by implanting the end of a severed peripheral nerve into a small, autologous free muscle graft. After reinnervation, electrodes implanted into RPNIs record highly specific and anatomically consistent EMG signals, which remain stable, allowing for precise control of individual fingers in humans for up to one year without requiring recalibration [6]. Previous work in able-bodied non-human primates has shown accurate tracking of digits, suggesting that control is intuitive as well as precise [7]. In this study, two participants with transradial amputations had bipolar recording electrodes surgically implanted into RPNIs and residual forearm muscles. The high-quality EMG signals recorded from the implants allowed a low latency pattern recognition system to predict individual finger movements and grasps in a virtual reality environment and during preliminary functional testing with a commercially available prosthetic hand. The high speed classifier also predicted movements in combination with a regression algorithm to provide 2 degree of freedom (DOF) position control of the index and middle-ring-small (MRS) fingers of a virtual hand to complete a dextrous target matching task.

METHODS

Two patients with transradial amputations, P1 and P2, had RPNIs surgically created on each of the median, ulnar, and radial nerves. P1 had one RPNI created on each nerve, while P2 had two RPNIs surgically created on the ulnar nerve, which had been subdivided into two fascicles, and one RPNI created on each of the median and radial nerves. Both participants provided written and informed consent and this study was approved by the Institutional Review Board at the University of Michigan. Eight pairs of bipolar electrodes (Synapse Biomedical, Oberlin, OH) were implanted into the ulnar and median RPNIs for both subjects as well as six and five residual muscles for P1 and P2, respectively. Although wrist movements were not a focus of this study, each subject had one electrode pair implanted in flexor carpi radialis (FCR). The remaining residual muscles were selected to target thumb, index, and small finger flexion and extension.

For 7 total experiment sessions, a Matlab xPC (Mathworks, Natick, MA) decoded EMG in real-time and controlled virtual [8] and physical (DEKA, Manchester, NH) prosthetic hands. Controllers were calibrated by having participants mimic 5-10 movement repetitions with their phantom limb while seated at a table. Training for virtual posture matching and functional grasps instructed participants to make discrete holds as opposed to gradual and intermediate movements for the continuous motor task. A Hidden Markov Model (HMM) was fit to training data and modelled transitions between latent states [9]. The underlying classifier, features, and processing windows were selected from other studies [6,7]. P1 performed preliminary functional tests where HMM output was directly mapped to pinch (Pi), point (Po), and hand close (HC), while rest (Re) predictions opened the DEKA hand (Figure 1). P1 and P2 also performed a pilot test that required them to precisely move the index and MRS fingers of a virtual hand to target positions (Figure 2). The controller for this task was a switching Kalman filter (KF) [10] with regression coefficients fit according to previous work [6,7] and an HMM to distinguish flexion of individual finger groups along with flexion and extension of both. Three performance metrics were evaluated per trial: acquisition time was the total time excluding a hold period, orbiting time was the time spent stabilizing around the target position, and path efficiency was defined as the distance ratio of a perfect 2D path to the actual path including orbiting (Table 1). These metrics were specifically chosen to evaluate the fine motor ability afforded by the intramuscular signals and controller.

RESULTS

P2 controlled a virtual hand in real-time to match a cue hand and select 7 postures: thumb, index, ring, and small finger flexion, fist, finger abduction, and rest. The HMM issued an incorrect prediction transitioning to the cue on 8.64% of trials, however P2 was able to quickly recover from these errors and hold the cued posture for 1 second with a 100% success rate. P2's average latency between the onset of new EMG activity and a successful hold was 311±31.2ms. Total trial time including reaction and hold was 1.73±0.03s on average (mean±s.e.m, n=73 trials across 3 sessions).

P1 controlled the DEKA hand with a HMM and completed a reach and place task (Figure 1) with an average time of 18.39±2.77s (mean±s.t.d, n=5 trials). Real-time accuracy was calculated by comparing the instructed grips for interacting with each object to the HMM commands output to the hand. Most misclassifications occurred when using the point grip during the button press, which was found to be a result of moderate index flexor activation.

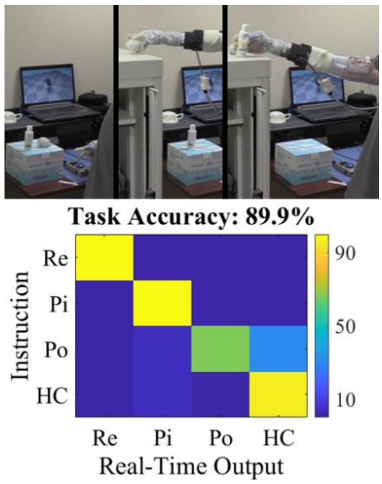

Figure 1: P1 performing the reach and place task which required three separate grips: point to press a timer button, pinch to move a ball, and hand close to move a bottle. P1 was instructed to start the timer, place both items on the shelf, bring the items back to the table, and stop the timer. Real-time accuracy was calculated across 5 trials.

P1 and P2 both used the switching KF to perform the dextrous 2 DOF target matching task which evaluated fine motor performance. The virtual task required them to navigate to 9 precise finger positions and remain within a tolerance window of ±13% flexion for 0.5-1s. Both subjects completed the task with success rates of 99%. On average P2 could not manage to move to target positions as directly as P1, evidenced by lower path efficiency and higher acquisition times despite comparable orbiting times (Table 1). This indicates that the P1 was better able to use the control algorithm to independently make fine movements.

Table 1: Dextrous 2 DOF Target Task Metrics

Participant	**Successful Trials (n)**	**Metric (mean±s.e.m.)**		
		Acquisition Time (ms)	***Orbiting Time (ms)***	***Path Efficiency (%)***
P1	100	871.8±77.4	190.5±72.0	74.2±2.5
P2	109	1025.7±82.3	141.2±51.3	63.2±2.5

Figure 2: P2 performing the dextrous 2 DOF target matching task by simultaneously and precisely matching the positions of the virtual index and MRS fingers (*grey*) which she had position control over to their cued positions (*blue*). The cue turned green to indicate successful positioning of the fingers.

DISCUSSION

This study demonstrated that electrodes surgically implanted into residual muscles and RPNIs allow pattern recognition of individual finger movements and functional grasps. The HMM did not require lengthy integration windows, allowing P2 to quickly recover from errors and complete the 7 posture virtual task with low average latency and a perfect success rate. P1 was also able to use the HMM and the DEKA hand to perform a task that required interacting with objects at multiple elevations. The common misclassification noticed during this preliminary functional test could be the result of subconscious muscle activity to stiffen the index finger for a button press. Similar phenomena have been noted by other groups and a variety of strategies exist to prevent such errors in future work [2,4]. The HMM implementation used a Naïve Bayes classifier to model latent states. However, it is likely that many classifiers could provide comparable performance due to the high amplitude and anatomical specificity of intramuscular EMG [6].

P1 and P2 also piloted a 2 DOF controller and performed a dextrous target matching task with similar near perfect success rates. P2's slightly lower average orbiting time may have been an artefact of a lower required hold time than P1. The larger discrepancies in other metrics suggest that for P2 either the HMM was not as effective in suppressing undesired movements or the movement distinctions were less intuitive. Strategies that blend trajectories of a switching KF may mitigate these issues [11]. The 2 DOF target task assessed fine motor control of independent finger groups. With commercial myoelectric systems using surface EMG, users rely on features of prosthetic hands such as compliant joints or internal controllers to substitute fine actuation for a gross motor command. Providing users with direct fine motor control of their prostheses will increase confidence over a broader range of activities, particularly as research in sensory feedback mechanisms progresses. Long term goals of this research are to increase the number of DOF and precision of finger control and incorporate precise control of wrist movements into a fully dextrous controller.

ACKNOWLEDGEMENTS

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ROBUSTNESS OF FREQUENCY DIVISION TECHNIQUE IN A SIMULTANEOUS AND PROPORTIONAL MYOELECTRIC CONTROL SCHEME

Ashirbad Pradhan[1], Ning Jiang[1], Wendy Hill[2], Victoria Chester[3], Usha Kuruganti[3]

1. *Engineering eBionics Lab, Department of Systems Design Engineering, University of Waterloo*
2. *Institute of Biomedical Engineering, University of New Brunswick*
3. *Andrew and Marjorie McCain Human Performance Laboratory, Faculty of Kinesiology, University of New Brunswick.*

ABSTRACT

It is important for myoelectric control schemes to be robust to various non-stationarities in electromyography (EMG) signal such as unintended activations and contraction level variations. In order to address this limitation, the present study compared performance measures of two EMG processing pipelines with two filtering techniques: frequency division technique (FDT) and standard bandpass processing (Bandpass) in a simultaneous and proportional myoelectric control (SPEC) scheme for two contraction levels (medium and high). Twenty able-bodied participants (14 males and 6 females, age 23.4 ± 3.0) performed wrist movements (flexion/extension, rotations and combined movements) in two degrees-of freedom (DOF) virtual tasks. FDT had a mean completion rate (CR) of 95.33%, which was significantly higher than the SPB technique with a CR of 64.08% ($p<0.001$). FDT method performed significantly better in all other performance indices in at least one movement type. Furthermore, there was no significant difference in the performance of FDT between medium and high contraction levels, while there were such differences for bandpass filtering. This study showed that FDT is advantageous in regression based online myoelectric control as it generates a more accurate, robust and contraction level invariant scheme for performing prosthetic hand movements. This study is the first to use frequency-based features with a SPEC scheme and shows promise for more intuitive prosthetic devices.

INTRODUCTION

Myoelectric prostheses use EMG signals for performing prosthetic functions. Conventional control of a myoelectric prosthesis involves mapping the amplitude of EMG signals to the desired prosthetic function. Challenges with the direct control scheme such as EMG crosstalk have led to the use of pattern recognition (PR), a machine learning approach that classifies EMG features to activate different prosthetic functions [1]. Currently, the state-of-the-art PR technique uses linear discriminant analysis (LDA) classifiers applied to a set of time domain (TD) features [2]. However, PR techniques only allow control of one DOF at a given time (sequential control) which is contrary to the natural control flow of the neuromuscular system. In order to achieve a more natural hand movement, simultaneous rather than sequential control is more desirable. Recently researchers have explored regression techniques, which allow for simultaneous and proportional control of the prosthesis [3, 4]. It has been found that linear regression (LR) performed superior to PR in an online closed loop setup [4]. The promising results of regression techniques has warranted further research to improve control of current prosthesis.

However, regression and PR techniques demonstrate relatively poor performance in real-world conditions due to the non-stationarities in EMG patterns and the noise introduced from different sources [5]. These variations or the non-stationaries in EMG may be caused by several factors including variations in training muscle contraction levels [6] and activation of an undesired degree of freedom [7, 8] are critical. One filtering approach using a frequency division technique (FDT) was proposed to increase varying contraction levels in PR-based myoelectric control [9], and this approach was demonstrated in a closed-loop online PR experiment [10], where. the control scheme with the FDT filter was found to be robust against varying levels of training contraction and it performed significantly better than the traditional band-pass technique. Further research with the FDT filtering on simultaneous and proportional myoelectric control (SPEC) scheme paired with FDT is warranted to corroborate findings in the PR-based myoelectric control scheme. Therefore, the purpose of this study was to compare the performance of the FDT and the traditional bandpass processing on a linear regression (LR) based online myoelectric control scheme while intact subjects completing virtual tasks. This study also examined the effects of varying training contraction level on the performance of the FDT based myoelectric control scheme to determine its robustness against force variation.

METHODS

Frequency Division Technique (FDT)

The FDT directly calculates the spectral power of various frequency bands of sEMG using discrete Fourier transform (DFT) by dividing the full bandwidth of sEMG signals into L segments. For the ith segment, let $f_{i,1}$ and $f_{i,ni}$ denote the frequency values of the two endpoints. The feature is defined as

$$DFT_i = F\left[\sum_{j=1}^{n_i}\left|X\left(f_{i,j}\right)\right|\right],\ i=1,2,\ldots L \qquad (1)$$

where, $X(\cdot)$ denotes the magnitude of the FFT spectrum, F denotes a non-linear smoothing function. In the current study, F is the root operator is used with a value of 2/3. The whole frequency band of EMG (20-450 Hz) is subdivided into six (L=6) equi-width frequency bands (20-92, 92-163,163-235,235-307,307-378, and 378-450 Hz) [10].

Protocol

Twenty intact-limbed participants (6 females and 14 males) with a mean age of 23.4±3.0 years participated in the study. The study was approved by the University Research Ethics Board (REB 2018-079). The participants were asked to sit on a chair in an upright position with both of their upper limbs in a resting position. They faced a computer screen, at an approximate distance of 75 cm. Eight equally spaced (19 mm inter electrode distance) bipolar electrodes (Duotrodes, Myontronics, Inc) were placed at approximately 1/3 distal measured from the olecranon process to the styloid process of the ulna to cover the circumference of the forearm. A commercial wireless biosignal amplifier (Trentadue, OT Bioelettronica, Italy), sampled at 1000 Hz, was used to transmit the signals. The dominant forearm was used for the electrode placement.

Feature Extraction and Testing

The surface EMG signals were processed initially using the common averaging method [10]. This was followed by two filtering techniques for two separate analyses, the band-pass filtering and FDT. The Bandpass filtering involved applying a bandpass filter (second order, Butterworth) from 20 Hz to 450 Hz followed the TD feature set extraction [10]. For FDT, the signals from each channel were divided into specific frequency sub-bands. LR was used for the simultaneous and proportional scheme. The outcomes of the regression model were mapped to the virtual task.

The experimental testing session consisted of two phases: 1) calibration phase and 2) control phase. The window size for processing was set to150 ms and the regression models provided an output every 50 ms. The calibration phase involved training a regression model using EMG signals with position labels of the cursor during wrist flexion/extension (DOF1) and hand pronation/supination (DOF2). In the calibration phase, the participants were instructed to follow the position of a cursor on a screen. In the training phase, the subjects performed two contraction levels: the wrist movements at the normal contraction level, *i.e.* 'train-medium', and wrist movements at a strenuous contraction level, *i.e.* 'train-high'.

From the data acquired in the training phase, a LR model were generated for each of the combination of the two contraction levels, *i.e. train-high* and *train-medium* and two filtering techniques: Bandpass, and FDT, resulting in four experimental sets in the control phase: medium-Bandpass, medium-FDT, high-Bandpass and high-FDT. In the subsequent control phase, the participants performed goal-directed tasks using the four LR models in a random order [10]. In each experimental session, 20 targets from each type of task group, termed type I, type II, and type III at

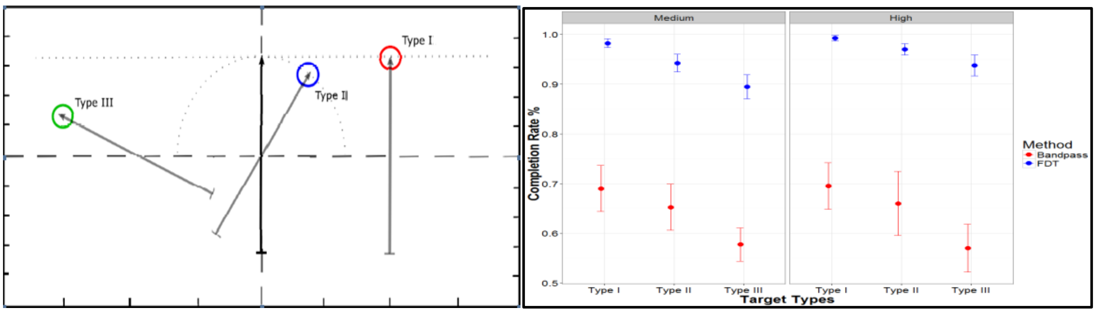

Fig. 1. Left: Goal oriented tasks: type I (flexion/extension DOF only), type II (pronation/supination DOF only) and type III (combination of flexion/extension DOF and pronation/supination DOF). The grey arrow represents the desired position for the completion of the tasks. Right: Mean CR for varying contraction levels (train-medium and train-high) and different processing methods (Bandpass and FDT) for the three types of targets (type I, type II and type III). The error bars represent the standard error.

different locations were provided on the screen (Fig. 2). The targets in type I only require the activation of wrist flexion/extension (DOF1), targets in type II require activation of wrist supination/pronation (DOF2), and targets in type III requires activation of both DOFs. The participants were instructed to place the tip of the arrow in the targets. Instead of sequential articulation of each DOF as in a PR-based control scheme, a simultaneous articulation of both DOFs was used. To measure the performance of these tasks, the performance indices used were: 1) completion rate (CR), the ratio of number of successfully completed task to the total number of tasks in percentage 2) time to reach (T2R), time taken to reach a target in seconds 3) throughput (TP) ratio of task difficulty and task completion time in bits/s and 4) near miss (NM), number of times the cursor enters the target but exits before the completion of 300 ms.

Kruskal Wallis (non-parametric test) was used to determine if the CR of the two filtering techniques were significantly different. Also, for the control participants repeated measures analysis of variance (ANOVA) was used to test for significant differences in mean performance indices (T2R, TP, NM) between FDT and Bandpass from successful trials. With significance resulting from the interaction of main factors the Bonferroni post hoc comparisons were performed to test significant differences in performance measures between FDT and Bandpass. For all the tests, level of significance was $p<0.05$. All the statistical tests were performed using RStudio 1.0. 136 (RStudio, Boston, MA).

RESULTS AND DISCUSSIONS

The mean CR of FDT was 95.33%, which was significantly higher ($p<0.001$) than Bandpass which had a mean CR of 64.08%. This indicates that FDT clearly outperforms the Bandpass. This was supported by the lower variability in CR for FDT compared to Bandpass, indicating less inter-subject variation. In addition, all participants performed equally well with FDT. The same training data was used to train both the processing/feature extraction methods. It was observed for most of the participants that while performing the Bandpass technique, the task arrow was unresponsive in at least one of four LR models. There was also frequent unwanted activation of the non-target DOF. For example, when an individual attempted a wrist extension there was undesired activation of supination as well. On the contrary, the FDT (CR>95%) was robust to unwanted activations and provided a more efficient control scheme. These activations have been briefly discussed by previous regression studies [7, 8], but there has been no detailed analysis on unwanted activations and it is crucial for further studies to research these non-stationarities and mechanisms of addressing them.

The mean T2R was significantly lower ($p<0.001$) for two types of targets (I and III) with FDT than Bandpass (Fig. 3). The mean TP was significantly higher ($p<0.001$) for two types (I and III) with FDT than Bandpass (Fig. 3). The mean NM of only type I target was significantly lower ($p<0.001$) for FDT. The Bandpass performed significantly better ($p<0.001$) than FDT for type II targets. A lower NM implies a more accurate position control. For FDT, the T2R and TP values suggested that the participants performed type I (horizontal only) and III (horizontal and rotation) tasks more easily and at a faster rate. Also, for both techniques, the variability was observed to be consistent for T2R, TP and NM (Fig. 3) suggesting that the participants had equal performance for all target types and across contraction levels. The overall TP and T2R values found in this research were comparable to previous study [10], however the NM was found to be higher. A possible explanation for higher NM is for some participants, the task arrow was unstable at higher pronation and supination angles, thus the participant had to hold it for longer increasing the NM. The mean NM was still low enough to allow real time control and the participants were able to complete tasks.

It was found that there were no significant differences in the mean values of any of the performance measures (CR, TP, T2R, and NM) between the train-medium and the train-high runs for FDT. For CR, the variability was lower

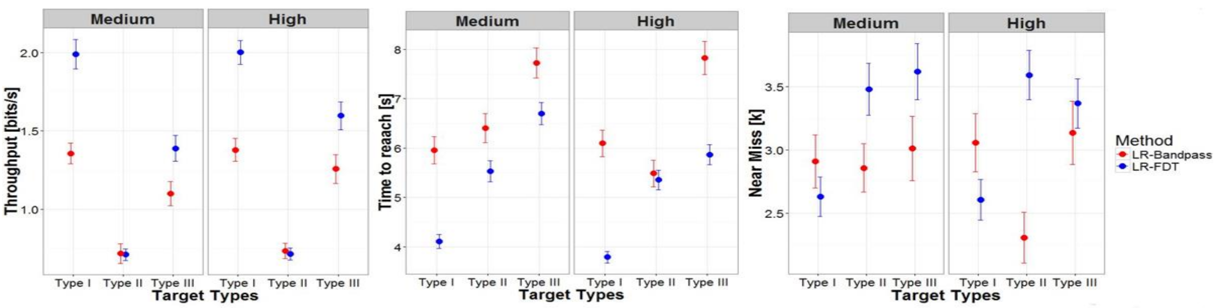

Fig. 3. (From left to right) Mean TP, T2R and NM values for varying contraction levels (train-medium and train-high) and different processing methods (Bandpass and FDT) for the three types of targets (type I, II and III). The error bars represent the standard error.

for FDT than for Bandpass (Fig. 2). For T2R, TP and NM, the variability was found to be consistent across contraction levels for both FDT and Bandpass (Fig. 3). This demonstrates that the performance of FDT is robust to contraction level variations while training. This observation agreed with the findings in [10], which used PR-based methods with FDT and found no difference between performance measures of medium and high contraction level variations [10]. Previously it has also been found out that the power spectrum of some frequency bands are not affected by varying contraction levels [9]. For the testing phase, the participants could perform tasks with any contraction level (medium or high). This finding is very important as the participant's control is independent of the contraction level performed during the training. A freedom of performing movements at a desired contraction level without any performance degradation would be beneficial for the prosthesis users to complete daily living tasks with limited errors.

CONCLUSION

The results from this study suggest that the proposed FDT performs significantly better than the Bandpass method in a LR-based control scheme. Also, the FDT technique is less variant to changing contraction levels. The two processing methods compared in the study used time domain (TD) features and frequency domain (FD) features. Most research studies to date have used the TD feature set. Results found in this study are promising and suggest a need for further research using FD features. The findings of this study directly relate to the robustness of FDT as a myoelectric control scheme which is critical for clinically viable advanced prosthetic control. In another research work (currently under review), the FDT technique demonstrated higher completion rates for individuals with trans-radial amputations compared to the Bandpass. Robustness against these non-stationaries allows users the freedom to operate a prosthesis at their desired contraction levels and prevents erroneous prosthetic functions. Thus, FDT in SPEC control scheme promises greater accuracy, robustness to varying contraction levels, and is more intuitive.

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SIMULTANEOUS AND PROPORTIONAL DECODING OF STIFFNESS AND POSITION INTENTIONS FROM TWO SEMG CHANNELS FOR UL PROSTHETICS

P. Capsi-Morales[1,2], C. Piazza[1], M. G. Catalano[2], A. Bicchi[1,2] and G. Grioli[2]

[1]*Centro "E. Piaggio" and Dipartamento di Ingegneria dell'Informazione, University of Pisa Largo Lucio Lazzarino 1, 56127 Pisa, Italy.*

[2]*Istituto Italiano di Tecnologia, Via Morego 30, 16163 Genoa, Italy.*

ABSTRACT

To physically interact with a rich variety of environments and situation-oriented requirements, humans continuously adapt both the stiffness and the force of their limbs through antagonistic muscle coactivation. Reflecting this behaviour in prostheses may promote control naturalness and intuitiveness and, consequently, their acceptance in everyday life. We propose a method capable of a simultaneous and proportional decoding of position and stiffness intentions from two surface electro-myographic sensors placed over a pair of antagonistic muscles. First, the algorithm is validated and compared to existing control modalities. Then, the algorithm is implemented in a soft under-actuated prosthetic hand (SoftHand Pro). We investigated the feasibility of our approach in a preliminary study involving one prosthetic user. Our future goal is to evaluate the usability of the proposed approach executing a variety of tasks including physical social interaction with other subjects (see Figure 1). Our hypothesis is that variable stiffness could be a compromise between firm control and safe interaction.

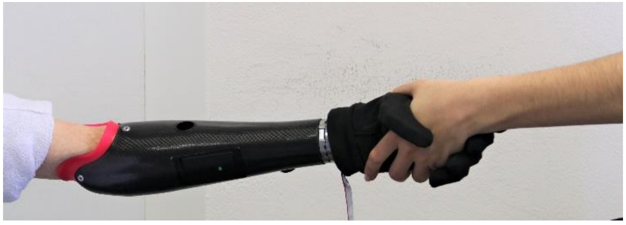

Figure 1: Physical social interaction between two subjects mediated by a soft robotic hand with variable stiffness control.

INTRODUCTION

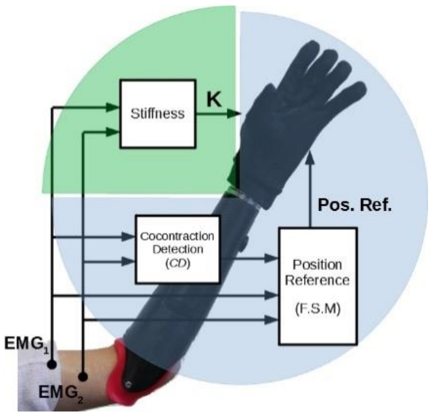

Figure 2: Block diagram of the proposed method: Three functional blocks decode stiffness (green colour) and position (blue colour) references from a pair of sEMG.

Artificial limbs are very valuable assets to restore some of the capabilities lost after an amputation. However, there is still a sharp separation between available functional devices and the real needs of prosthetic users [1]. Social interaction and safety are aspects that cannot be underestimated in prosthetics, especially in upper limb, due to the inherent interaction of the artificial hand with, not only the user, but also the rest of the world. Already in 1983, Hogan [2] suggested impedance control as the preferred paradigm for controlling prostheses, as it would provide the amputee with an essential component of the natural adaptative capability of humans, despite the severe sensory loss. Moreover, behavioural studies of postural limb control show that humans modulate joint stiffness to minimize the perturbing effects of external loads [3] and to improve limb stability and movement accuracy [4]. However, muscles stiffness regulation is not available in off-the-shelf prosthetic aids, neither its investigation is given, in literature, the space we believe it would deserve, both under the control [5] and mechatronics points of view.

We introduce a method for decoding an estimate of user's stiffness intention, based on cocontraction, which can be used simultaneously within a proportional velocity control framework, thanks to the inclusion of a custom Finite State Machine. The primary objective is to exploit cocontraction for a useful and intuitive increase of direct control robustness, a better decoding of patient's intentions and to enlarge prosthesis dexterity. The novel control was preliminary tested with one subject with limb loss, with encouraging results.

DECODING OF STIFFNESS & POSITION

It is well known that the position of a joint is defined by the equilibrium of the various muscles acting on it, together with external forces. However, the multiple action of antagonistic muscles, i.e. coactivation, defines the mechanical properties of the joint as well. In order to benefit from the intrinsic muscle stiffness regulation that humans have, we propose an algorithm capable to decode both position and stiffness intentions from two sEMG channels with the inclusion of coactivation.

A common method to estimate the level of coactivation of a pair of muscles is to correlate it to the weighted average of the level of activation of the two antagonistic muscles (e.g. as in [6]), as

$$K = C_1 EMG_1, C_2 EMG_2 \ . \tag{1}$$

Stiffness K can either increase due to involuntary reaction to external disturbances, voluntary cocontraction, or reciprocal muscle activation. Unfortunately, combining this estimate with traditional velocity control schemes, would have the inconvenience that pure cocontraction phenomena, unless perfectly symmetrical and synchronized (which never happens in practice), would be interpreted as either open or close commands, depending on which of the two EMG signals is observed overcoming its threshold first.

To prevent this issue, we observe that usually the level of the extensor muscle contraction is almost zero when closing, and the opposite happens when opening. Therefore, we can define an additional variable used for binary (true/false) detection of pure cocontraction, *CD*, as in

$$CD = \begin{cases} 0, & if\ min(C_1 EMG_1, C_2 EMG_2) \ < \ Th_{CD} \\ 1, & otherwise \end{cases} \tag{2}$$

where Th_{CD} is a suitable threshold value. Consequently, *CD* will be one only when cocontraction is intended, as both sEMG have a high level of activation, and null when only one of the two sEMG is above the threshold, indicating a motion intention.

Note that the calculation of *CD* and *K* are simultaneous and independent, thus the algorithm keeps generating commands of velocity and stiffness simultaneously. It is possible to observe that there is some correlation between motion commands and stiffness, since when the user contracts one muscle, e.g. to close the hand, it will always command a minimum level of stiffness, proportional to the minimum level of activation needed for the active muscle to overcome its threshold (see Figure 3), but this reflects the natural behaviour of muscles, being a desired and welcome effect.

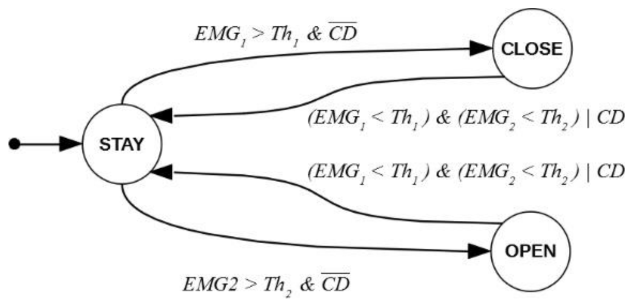

Figure 3: Finite States Machine used to refine the speed of the hand. Hand states are defined by circles, while guard conditions are written directly nearby the arrow connecting pre and post states. The starting state is STAY. *Th1* and *Th2* are the activation threshold for each channel to detect intention of movement.

Figure 3 shows the Finite State Machine that is ultimately used to discriminate the user's intention to modify the reference configuration (*RC*) - opening or closing the hand - or to hold it still. The definition of the hand *RC*, in analogy with typical velocity control frameworks, is updated as

$$\dot{RC} = \begin{cases} C_1 EMG_1, & if\ CLOSE \\ 0, & if\ STAY \\ C_2 EMG_2, & if\ OPEN \end{cases} \ . \tag{2}$$

ALGORITHM VALIDATION

We present an example of the behaviour of the proposed algorithm, and compare it to existing methods, highlighting the different interpretation of the user's intentions. Two EMG signals were collected and read into MATLAB Simulink (Mathworks, Inc) from a healthy subject (female, age 26). Two commercial sEMG sensors were used to get the signals (13E200=60, OttobockGmbH, Germany). The various control algorithms were run in Simulink.

The first panel of Fig. 5 shows the EMG activations, while the three remaining panels present the system responding to the following control modalities:

- PPC-PS: Proportional Position (used e.g. in [6]) with Proportional Stiffness;
- PVC-HS: Proportional Velocity (FSM 1) with High constant Stiffness;
- PVC-LS: Proportional Velocity (FSM 1) with Low constant Stiffness;
- PVC-PS: Proportional Velocity (FSM 2) with Proportional Stiffness.

where FSM 2 refers to the machine previously detailed (figure 3), while the FSM 1 does not consider CD as an input of its conditions.

The differences between the two variables related to muscle coactivation used in this work (*K* and *CD*) are observable in figure 4. *K* is presented in the second and fourth panel, proportional to muscle activation. *K* tends to rise both in the case of cocontraction and of pure contraction, making the detection of pure cocontraction phenomena hard. This leads to the introduction of *CD*, which is defined by the variable *cdet* represented by the grey area of the first panel. Comparing *cdet* to a simple threshold (the dashed black line), there is a clear categorization between cocontraction and other types of muscle activation.

Concerning position reference, in the case of proportional position control (second panel), based on the difference between EMG_1-EMG_2, we observe how cocontraction is interpreted in opposition to as intended, reducing the level of activation. This results in motion of the hand in a direction opposed to the desired one. In addition, although PPC is more reactive to muscle variations, the subject must keep the muscle active during all the time in order to maintain the hand closed. This is tiring both from a physical and mental point of view. Regarding the performance of the two FSM for PVC, FSM 1 already solves the problem of the tiredness, but is not able to understand pure cocontraction as an extra user's intention. As seen in the third panel, the FSM states does not correspond with the real intentions, not just closing the hand when cocontracting, but also not opening the hand in the first part, as no relaxing phase (STAY state) occurs before the opening intention (around t = 5s). On the contrary, FSM 2 understands correctly subject's intentions, closing and opening just when the proper muscle is active, and remaining in STAY state when nothing occurs or if the user cocontracts to increase the stiffness but not intend to change hand position. Although it is noticeable some very fast oscillations of the FSM 2 close to cocontractions, when just one of the EMG is active (e.g. shortly after 14s and before 22s), in practice these oscillations do not affect *RC* sensibly.

GRASP COMPLIANCE

The proposed method was implemented on a soft underactuated hand device, the SoftHand Pro [7]. To the best knowledge of the authors, this is the first experimental validation of impedance control in prosthetics hands performed by an amputee, indeed, previous works as [8] and [9] used healthy subjects only. We validate the feasibility of the control algorithm used by a prosthetic user with informed consent. The subject (female, age 37) has a congenital malformation at the trans-radial level in the left arm. She typically uses a cosmetic prosthesis but is well trained in control of standard myoelectric prostheses. We study the response of the system with regards to an external perturbation while using the four control algorithms presented in the algorithm validation section.

CONCLUSIONS

In order to achieve different desired behaviours in upper limb prosthetics for Activities of Daily Living and for social interactions, an alternative solution to the classical sEMG based control is explored. The proposed method includes stiffness modulation of the hand, proportional to muscle coactivation with a proportional velocity control of the hand configuration with the use of a Finite State Machine. The algorithm is preliminarily validated with a prosthetic user, comparing it with other conventional control modalities implemented in the SHP. Eventually, this concept could be implemented in other rigid prosthetic hands, where differences between modalities could be even more visible/useful, as their only compliance can be given by the motor impedance.

Preliminary results evidence a better understanding of user's intentions through the inclusion of cocontraction on the control algorithm. Different performances are observed among control strategies studied, which could influence subjects' perception. For the moment, the user underlined the lack of confidence and difficulties to command the hand when using the Low constant Stiffness control (LS), because of the lack of reactivity, and Proportional Position Control (PPC), because of the amount of cognitive load required.

Our objective is to explore the perceived function of variable stiffness control compared to strategies with a constant stiffness value. User's preferences will be assessed by the System Usability Scale (SUS) [10] after performing a set of tasks without having any information about the control implemented in the prosthetic hand. Among these tasks, one- and two-handed object manipulation are included together with self-interaction and social interaction with 12 able-bodied volunteers. Volunteers reactions will be also collected and analysed.

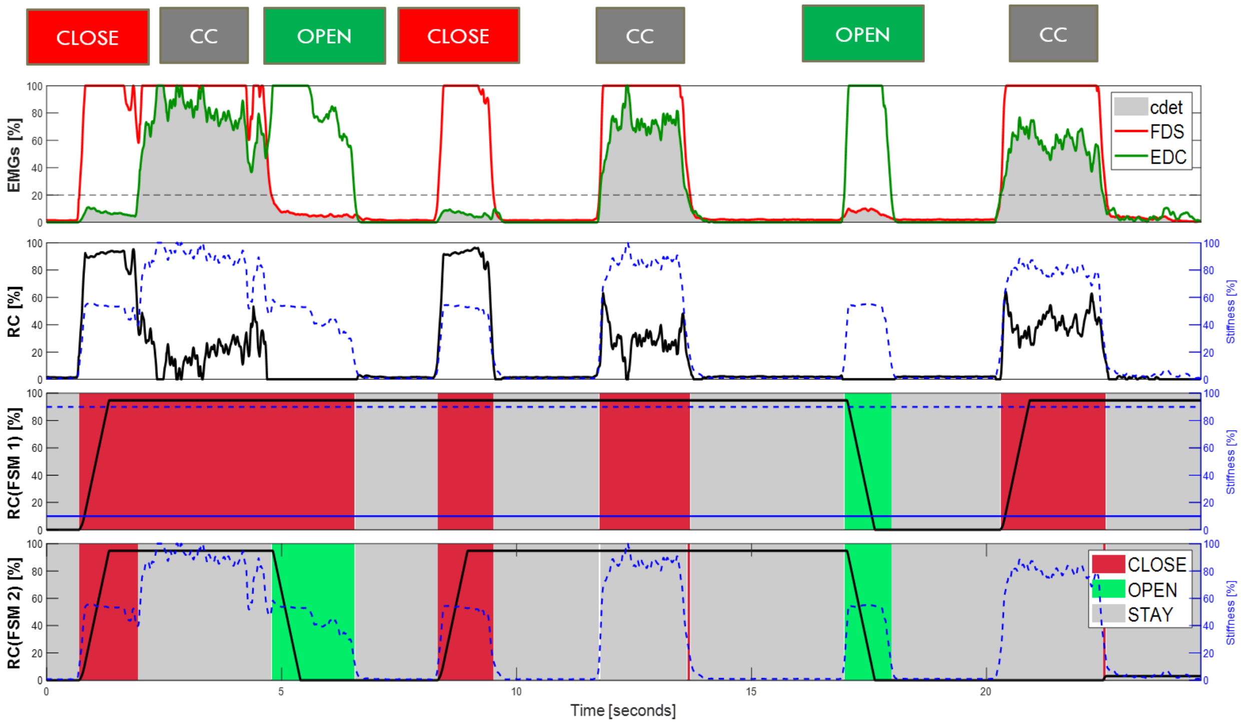

Figure 4: Experimental comparison of different control strategies responding to the same muscle activation. The first panel presents the muscle activation from a healthy subject performing a series of intentions described in coloured boxes in the upper part of the graph. The variable $cdet = min(C_1EMG_1, C_2EMG_2)$ defines *CD* for the FSM 2. Second panel shows the outputs of PPC-PS, third panel the overlapped outputs of PVC-LS and PVC-HS, while fourth panel reports the outputs of PVC-PS. Black line (left y-axis) reports RC while blue line (right y-axis) reports stiffness reference. In the third panel, both high and low constant stiffness are represented, HS is outlined with a blue dashed line, while LS with a blue continuous line. All quantities are normalized. In the bottom two cases, where FSM are employed, colours are used to represent the state (detected intention of the user) in each moment. Red areas correspond to when closing is understood by the algorithm, green when opening, and grey when the hand keeps the previous position.

ACKNOWLEDGEMENTS

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TRAJECTORY CONTROL FOR A MYOELECTRIC PROSTHETIC WRIST

Yuri Gloumakov, Joao Bimbo, Aaron M. Dollar

Yale University

ABSTRACT

We present a novel method for controlling a myoelectric prosthetic wrist. Five multiple degree-of-freedom (DOF) wrist trajectories are obtained from healthy participants that performed tasks that span the range of Activities of Daily Living (ADL) using dimensionality reduction and unsupervised machine learning techniques. The efficacy of these motions is tested as part of a pilot study where a participant used a simulated wrist device controlled using two-site surface electromyography (sEMG); two trajectories were tested in an immersive virtual reality. Novel wrist control has been demonstrated to be more intuitive to use and appears more natural while limiting the amount of body compensation.

INTRODUCTION

Orienting the hand has been shown to be as important as finger dexterity in aiding us perform Activities of Daily Living (ADL) [1]. Prosthetic devices featuring a wrist, however, have either only 1 or 2 degrees-of-freedom (DOF), largely due to a lack of intuitive control associated with orienting a hand in 3-DOF rotation space while operating each orthogonal DOF independently. Our work focuses on developing an intuitive control strategy for a 3-DOF prosthetic wrist device by taking advantage of joint angle synergies and identifying predefined wrist orientation trajectories that do not require users to independently control each DOF.

Synergies have been identified across different joints in the human body [2], and have been demonstrated to be effective in prosthesis use [3]. Some joints are also predictably coupled [4]. We found inspiration in these findings and identified sets of predefined full arm shoulder-elbow-wrist trajectories using unsupervised machine learning techniques that clustered whole wrist movements into defined sets [5]. The arm movements corresponded to healthy individuals performing a comprehensive set of activities of daily living (ADL). We implement a similar approach to identifying clusters of wrist movements, following with an averaging algorithm to obtain a small, yet representative, set of wrist trajectories.

Virtual Reality (VR) has been used across many domains dealing with the human hand. It can be a valuable tool for training the use of myoelectric prosthesis [6], and can be truly immersive; demonstrated through its capability to treat phantom limb [7]. We make use of advances made in VR technology to demonstrate the capacity of the proposed wrist trajectory control to be a practical approach to operating all 3-DOF of a prosthetic wrist.

METHODS

<u>Wrist Trajectories</u>

We obtained a set of representative wrist trajectories through a series of dimensionality reduction techniques. We first collected 12 healthy subjects (age 24-71) performing ADL using motion capture; 12 Bonita Vicon cameras tracked markers placed around the subjects' forearm and hand. The set of ADL were inspired by work done on upper-limb rehabilitation and prosthesis use evaluation [8], and include the following: drinking from cup or mug placed in various locations, transferring a suitcase or a box, reaching to a can overhead, pouring from a cup, eating with a fork or spoon, reaching to the axilla, and reaching to the back pocket; listed in more detail in our previous work [5].

Joint angles were extracted from marker data and clustered using Hierarchical Clustering with Ward's Distance measure, using dynamic time warping (DTW) to measure the similarity between motions. The number of clusters was identified using the L method. Each cluster was averaged using DTW barycenter averaging (DBA) to distil the large set of motions to a small set of representative wrist trajectories. This study protocol was approved by Yale University Institutional Review Board, HSC# 1610018511.

Control Modes

Participants completed the series of tasks using two types of wrist control: sequential control, and the proposed novel trajectory control. Sequential control interpreted the flexion sensor as driving the wrist along the positive angle direction while the other sensor drove the wrist in the opposite direction at constant speed. A co-contraction cycled the mode from *pronation-supination* to *flexion-extension* to *ulnar-radial deviation*, with pronation, flexion, and ulnar deviation being the positive directions.

The identified wrist trajectories are implemented in our proposed trajectory control. In this setup, the flexion sEMG sensor drove the wrist forward along a selected trajectory, while the extension sensor drove the wrist backwards along the trajectory at a constant speed. A co-contraction results in the cycling between the five trajectories. Each of the trajectory control modes have a defined start and end point. Therefore, even for sequential control conditions, the wrist began in the same orientation as the trajectory control.

Control Input

We used HTC Vive for both the head tracking and for the head mounted display (HMD). The participant's forearm was tracked and displayed within the virtual environment (VE), implemented in Unity, to provide a point of reference for the hand orientation. This was done using Vicon to track markers placed around the forearm and streamed to Unity. To control the virtual hand, the participant's forearm was also outfitted with two surface electromyography (sEMG) sensors, placed on the flexor and extensor muscle groups (see Figure 1), connected to an Arduino Uno. Sensor readings were translated to either *on* or *off* according to a calibrated threshold value.

(a) (b)

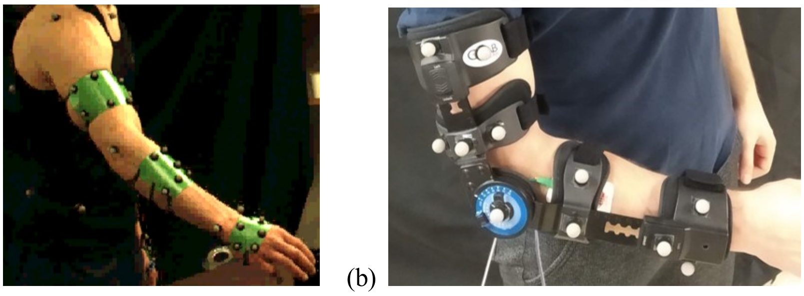

Figure 1: (a) Marker set used to collect healthy arms motions that were then used to generate the wrist trajectories. (b) The elbow brace was used to maintain the reflective marker arrangement, such that the virtual forearm and humerus segments are automatically detected and displayed within VR. The brace's range of motion was set to maximum and was not used to limit the elbow motion itself. sEMG sensors placed over the skin around the forearm can be seen underneath the elbow brace.

Pilot Study Procedure

In this pilot study, one healthy right-handed participant (male, age 28) performed two tasks related to ADL in VR by attempting to align the end effector with the desired goal. The subject did not have any visual or motion impairment and was comfortable using VR. Tasks included in this pilot study are described in more detail in Table A. Because each trajectory control mode corresponds to a specific task, these were included in the table for reference. Only two tasks were tested in this pilot, therefore only trajectories (4) and (3) (see Figure 2 for detail) were used, for reaching to the cup and pouring with the cup, respectively. Prior to each task, the participant was given ample time to practice and develop a strategy that they're comfortable with using during the task recording; the purpose was simulate the performance likely achieved by an experienced user. For tasks involving object transfer, objects were automatically placed within the hand.

Table 1: Pilot tasks

Task	Task description	Corresponding wrist trajectory
Reach to cup	Standing, starting with the hand by the side, reach to the cup on the table	(4) supination/extension
Pour from cup	Sitting, transfer the cup from the table to the pouring location and orientation	(3) supination/flexion

Evaluation

The participant's performance can be assessed in various ways. Because the goal is to improve prosthesis use in the real world, we wanted to focus on the time it takes to complete a task and the motion cosmesis. The participant also provided feedback and helped guide our interpretation of his performance. While cognitive effort to control the prosthesis was not directly measured, it may be inferred from the time measurements.

RESULTS

Wrist trajectories were obtained through averaging each of the motion clusters. Although each consists of a 3-DOF wrist rotation, they can be better described according to the dominant DOFs as follows: (1) *supination/ulnar deviation* (2) *flexion/ulnar deviation* (3) *supination/flexion* (4) *supination/extension* (5) *extension-ulnar deviation*, as seen in Figure 2. Two of these wrist trajectories, (3) and (4), were used in the pilot study.

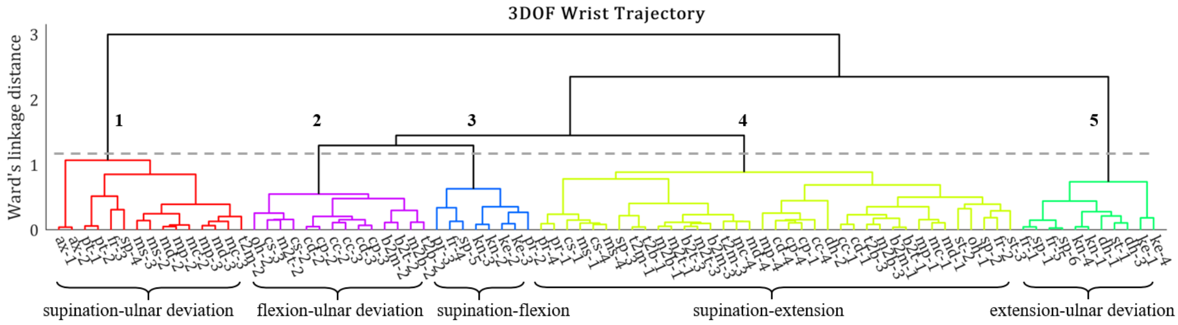

Figure 2: Hierarchical clustering results. A horizontal cut segmented the dendrogram into five clusters of motion. A descriptive label is included for each cluster.

Recorded wrist joint motion trajectories for each of the trials are displayed in Figure 3. Motions were segmented according to when the participant's hand began to move and when the target end effector position and orientation, was reached.

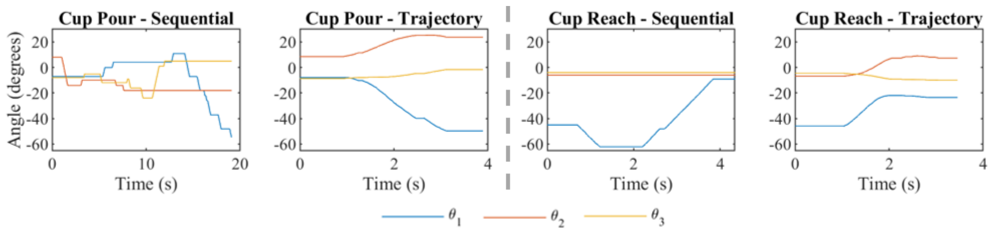

Figure 3: The 3-DOF wrist joint angle trajectories are displayed for each trial. θ_1, θ_2, and θ_3 correspond to pronation, flexion, and ulnar deviation respectively. The left two plots correspond to the cup pouring task under the two different control strategies, sequential and trajectory control, while the right two images correspond to the cup reaching task. Wrist rotation did not necessarily begin when the hand started to move.

The participant was able to complete both tasks faster using trajectory control. Sequential control for the cup pouring task took significantly longer than when using trajectory control, while the times were much closer for the cup reaching task. This is likely because the task required switching between the different joint angles, which can be challenging, or even confusing, for the user. The cup reaching task did not require switching between the different DOF, and supination alone was sufficient.

Wrist motions appeared more naturally under trajectory control. This is largely due to the lack of access to all 3-DOF of the wrist during sequential control, as is evident in Figure 3. Without haptic feedback, the user appeared to be looking down at their simulated device. This was exacerbated when multiple mode switching was required, such as for the cup pouring task with sequential control. Trajectory control for both tasks did not require mode switching, since a single mode, corresponding to the respective task, was sufficient.

DISCUSSION

In this study we were able to gain significant insight into our proposed wrist trajectory control that encourages further investigation. In this preliminary study, trajectory control has been demonstrated to be a superior alternative to sequential control, despite limiting users to specific wrist orientations. Findings further demonstrate the capacity of joint synergies to simplify control. Trajectories appeared to generalize well to the tasks, without requiring the user to compensate with their residual limb or torso.

During the experiment, when using sequential control, the participant generally relied on fewer DOF than were available. This was likely the easiest way to control the wrist without having to repetitively switch between DOF. This showcases the benefits of trajectory control whereby all 3-DOF of the wrist are at use while maintain a simple and intuitive control strategy.

We must also acknowledge that there were learning differences between the two control strategies. While sequential control would task users to learn the correct order of rotations, trajectory control requires a memorization of which tasks belong to which motion control. In the future, training time and cognitive load will be addressed.

Using state of the art motion tracking, HMD, and control input, we believe this is the closest a simulation can get to testing prosthesis without using the actual prosthetic device. Innovations in this field have the potential to streamline prosthesis design iterations, prosthesis training, and rehabilitation [9], [10]. However, there are certain drawbacks that need to be addressed in the future in order to fully bridge the gap between simulation and reality. These include adding haptic feedback, inertia, wider field of view and resolution in the HMD, and improving the realism of the virtual environment design.

In future iterations of this experiment we will recruit additional subjects and expand on the tasks. We will also include alternative state of the art control strategies, such as enabling participants to simultaneously control DOF. Positive and negative controls will be included as well, corresponding to tracking the users' hand while unrestricted and fully restricted, respectively.

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Track: Myoelectric Control and Sensory Feedback Implementations

DEMONSTRATION OF AN OPTOGENETIC NEURONAL CONTROL INTERFACE

Arjun K. Fontaine[1], Jacob Segil[3], John H. Caldwell[2], Richard F. Weir[1]

Departments of [1]Bioengineering and [2]Cell and Developmental Biology,
University of Colorado Denver|Anschutz Medical Campus, Aurora, Colorado, USA
[3]Engineering Plus Program, University of Colorado Boulder, Boulder, Colorado, USA

ABSTRACT

Improved nerve interface approaches are sought for prosthesis control and sensory feedback as well as visceral organ study/modulation. Optical approaches that can read-in and read-out neural activity have advantages over electrode-based systems in terms of selectivity and non-invasiveness. To address limitations of existing nerve interface designs, we present an optical approach capable of reading activity from individual nerve fibers using activity-dependent calcium transients. Here we demonstrate the feasibility of using activity-dependent calcium transients to a control prosthetic hand. This work provides a proof-of-concept for an optogenetic nerve interface demonstrating as it does our ability to read-out signals at the axonal scale in real-time and apply it to a devices control.

INTRODUCTION

We are developing a Bidirectional Optogenetic Neural Interface to read-in and/or read-out action potentials from a nerve with the goal of creating a neural interface that is selective yet minimally invasive to the nerve. There are significant drawbacks to current nerve interface approaches. They either lack specificity - they use nerve cuff electrodes, such as the Flat Interface Nerve Electrode (FINE) Array**[1]** that must sit on the outside of the nerve and measure signals originating inside the nerve bundle, or they involve penetrating the nerve with needle electrodes - such as Longitudinal Intrafascicular Electrodes (LIFE)**[2]** or the Utah Slant Array**[3]**. Penetrating electrodes tend to be hard and rigid, resulting in a stiffness mismatch that causes irritation and necrosis, decreasing longevity. Instead of using electrical interfacing with the nerve, we will use light activated ion channels (opsins) and fluorescence protein Ca^{2+} or voltage indicators that allow stimulation and recording of action potentials of specific afferent or efferent neurons using viral vector transfection. Our Optogenetic Neuronal Interface is based on a fiber optic coupled miniature two-photon microscope with electrowetting adaptive optics **[4-7].**

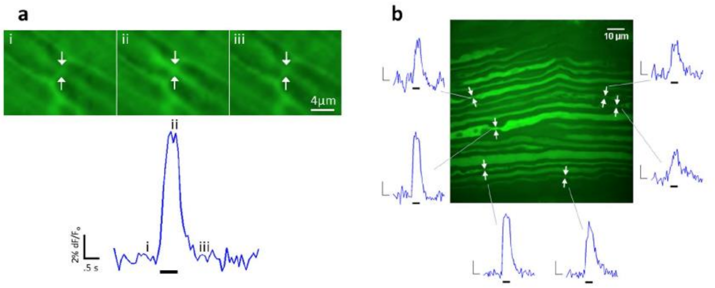

Figure 1: *Action potential elicited calcium signal detection in tibial nerve axon nodes of Ranvier [1].*

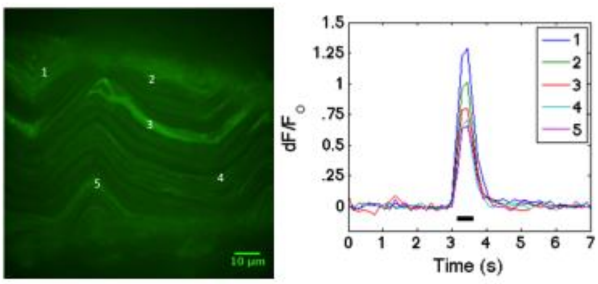

Figure 2: *Action potential elicited calcium signal detection in vagus nerve axons with GCaMP6f.*

The Bidirectional Optogenetic Neuronal Interface system is based on the principal of two-photon (TP) excitation**[8,9].** In TP excitation, a fluorophore is excited by short pulses of laser light. TP excitation offers intrinsic axial cross sectioning because the process only occurs at the focus of the objective lens. The technique offers resolutions of 175 nm lateral and 451 nm axial for 900 nm light focused with a 1.2 NA objective. This approach, when combined with a lateral scanning head, provides axon scale resolution that can be used to selectively interrogate an axon while excluding signals from the remaining tissue.

Peripheral nerve *read-out* of activity using calcium-sensitive fluorescent reporters: We have demonstrated read-out of genetically expressed activity-dependent calcium indicators, such as GCaMP6f, has been demonstrated in other work in vitro**[10]** [**Figure 1** & **Figure 2**]. We have also shown how a viral vector might be used as a mechanism for delivery of long-term optical protein expression in mouse neurons for optical read-out **[11]. Selective photo-stimulation (*read-in*) in nerve:** We have also demonstrated the ability to selectively read-in (or stimulate) to nerves optically [**Figure 3**].

Here we further demonstrate the feasibility of using optical approaches for prosthesis control by imaging the axonal fluorescence produced by action potentials travelling in an in vitro mouse nerve and using the change in image intensity to drive a prosthetic hand in real-time. This work provides a proof-of-concept for an optogenetic nerve interface demonstrating as it does our ability to read-out signals at the axonal scale in real-time and apply it to a devices control.

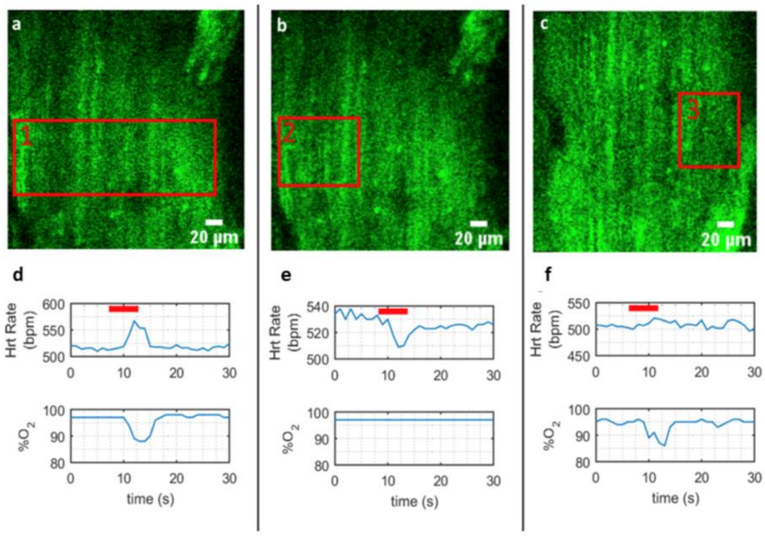

Figure 3: *Spatially selective photo-stimulation elicits differential vitals responses. (a-c) Regions (1-3) of 1040nm photo-stimulation within the cervical vagus nerve of an anesthetized ChAT-GCaMP6s mouse. (d-f) Corresponding vitals responses to photo-stimulation; region 1 elicits an increase in heart rate and a decrease in oxygen saturation; region 2 elicits a decrease in heart rate and no change in oxygen saturation; region 3 elicits a decrease in oxygen saturation. 1040nm stimulus was applied for 4 seconds with 20 ms pulses at 20 Hz.*

METHODS

Nerve Preparation: The sciatic nerve and its tibial nerve branch are excised from adult wild type mice and loaded from the tibial end with a synthetic calcium indicator (2 mM Calcium Green-1 Dextran, ex/em = 506/531 nm) dissolved in a buffer containing 130 mM KCl and 30 mM MOPS, pH 7.2 in *accordance with Supplementary Figure 1, Fontaine et al, 2017* ***[11]*** **(Figure 4) .** The tibial end is freshly cut in a zero-calcium buffer to ensure open axon cylinders before being suctioned into a tight-fit electrode with the dye buffer to facilitate longitudinal axonal dye-loading via diffusion and/or axoplasmic transport. The suction electrode on the tibial nerve also serves to record electrical activity within the nerve. The sciatic end of the nerve is drawn into a suction electrode for electrical stimulation of compound action potentials (CAPs). All experiments were performed in accordance with our Institutional Animal Care and Use Committee (IACUC) regulations and approved protocol.

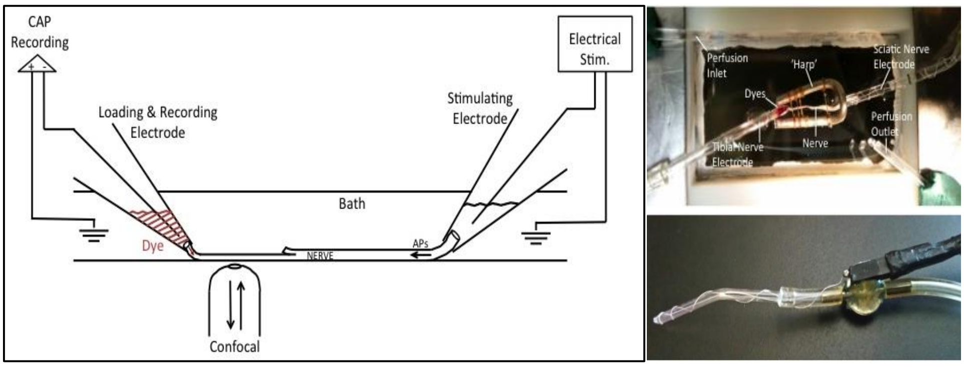

Figure 4: *Nerve dye-loading, electrophysiology & imaging configuration*

Electrophysiology: CAPs are generated and recorded throughout the experiment using 50 µs square pulses to confirm and monitor nerve viability. The stimulation voltage threshold for maximum CAP amplitude is determined. CAP amplitudes were monitored throughout the duration of the incubation period, to confirm stable nerve health.

Optical Imaging/Recording: Dye labeled axons were imaged in a region of nerve near the tibial recording electrode. The nerve was gently pressed to the optical glass of the chamber with low-tension silk strings attached to a small weight for imaging on an inverted microscope. Placement of the small 'harp-like' device did not affect the CAP. Fluorescence imaging was performed on a spinning disk confocal microscope (Intelligent Imaging Innovations, Marianas). A 515nm laser line was used to excite the Calcium Green-1. Pixels were binned (2x2) to improve the frame read-out time for fast imaging. To record calcium transients, time-lapse images were acquired at 12-20Hz (motor update rate), during which the nerve was stimulated by an electrical stimulator triggered via TTL pulses from the microscope. Fluorescence was imaged onto an EMCCD camera (Photometrics Evolve) through a 525/50nm emission filter. Images were collected with a 63X, 1.4NA oil-immersion objective lens. Photobleaching of the signal was kept minimal by the reduction of laser power and exposure, and any mild decay due to photobleaching was not removed.

***Prosthetic Hand Modification*:** The electronics in the original Bebionic v2 hand (RSL Steeper, UK) **(Figure 5)** were replaced with a custom motor controller system (Sigenics Inc., Chicago, IL) and included a central Arduino controller board and six satellite boards referred to as 'penny boards' (as they were the size of a penny). Each penny board was connected by a four-wire I2C bus with each board associated with an individual finger for finger flexion/extension with two for the thumb to drive flexion/ extension and abduction/ adduction. For velocity control motor commands indicating the speed and direction of motion for the driven finger were sent from a Matlab script to the Arduino (SparkFun Electronics, Boulder, CO) which converted the serial commands into I2C commands. Position encoder values from the prosthetic finger motor were recorded simultaneously and converted to joint angle measurements post hoc. For the Bebionic the fingers can flex from 0-95º and run at a max speed of about 2 rads/sec. For position control, desired finger position is sent over the I2C bus to the motor controller and a local on-board PID loop handles positioning of the finger.

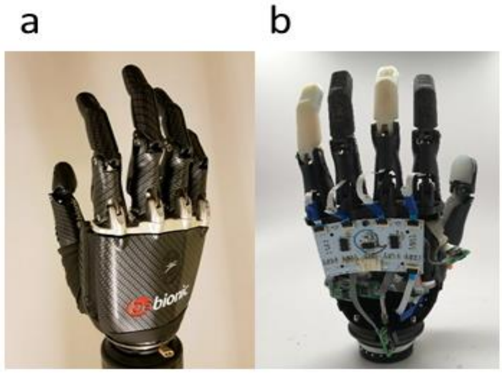

Figure 5*: (a) Commercially available Bebionic v2 hand (b) Modified Bebionic hand used for finger actuation experiments. Custom electronics were installed in order to control individual motors within the prosthesis. The Bebioinc has motor encoders that can measure finger position and be used in closed -loop control..*

Control Interface and Method: A standard laptop computer running SlideBook 6.0 software (Intelligent Imaging Innovations) took the raw time-lapse images from the microscope and sent them to a custom Matlab program (Mathworks, MA) which calculated the intensity of the region-of-interest (ROI) on the selected axon and based on the computed value sent commands to the motor controllers of the prosthetic hand via a serial link. A setup function in the Matlab script established the serial communication between the computer and the prosthetic hand. A second function received the time-lapse captures from SlideBook and translated the image data into an optical signal by averaging nodal ROI pixel intensities in each frame. The change-in-intensity is the control signal-of-interest. We see a baseline intensity for zero firing rate and a 15-18% for a firing rate of 125Hz. Since baseline is not constant, we set a threshold of 2%. This gives us our command signal range: for 0-125hz we expect a 2-18% dF/F which should map to 0-100% of our command signal for the motor. Initially we mapped the optical signal to the prosthetic finger velocity in an open-loop velocity control paradigm that is standard-of-care [**12**]. The hand was set up in a "cookie-crusher" configuration so single-site control could be used. In this case when the amplitude of the signal rises above the optical signal threshold the finger was driven in flexion at a speed proportional to the change-in-intensity. Velocity gains were adjusted to achieve a full range of motion.

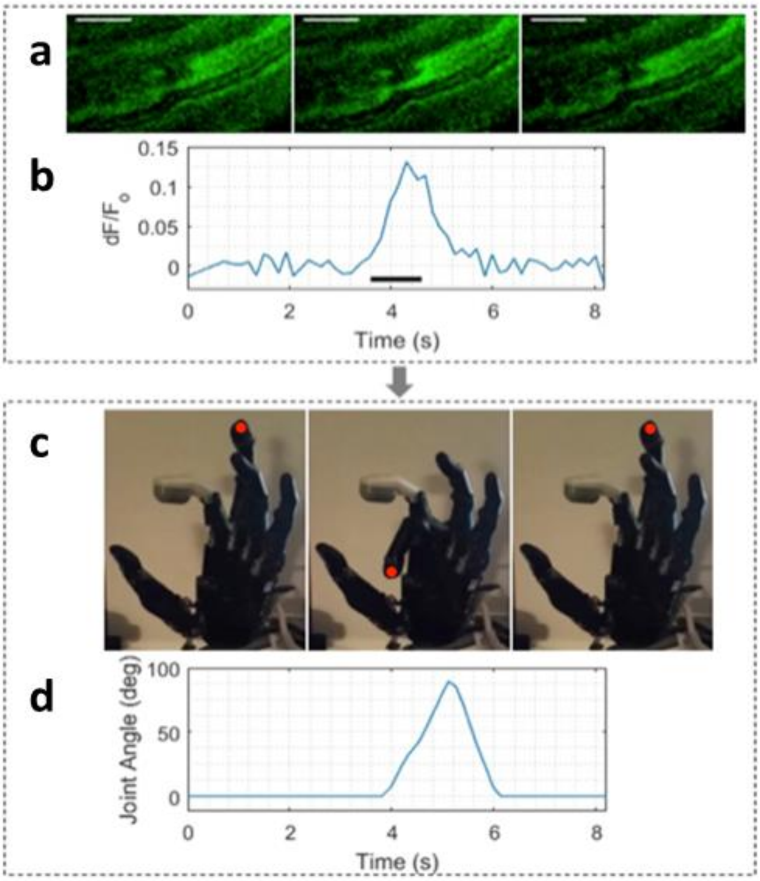

Figure 6*: Real-time prosthetic digit actuation by action potential evoked calcium fluorescence signal in a peripheral nerve axon. (a) Confocal images of a CalciumGreen-1-Dextran-loaded axon node-of-Ranvier used to control finger actuation, shown before, during and after the activity-induced fluorescent signal (scale bar 10μm). (b) Quantitative trace of the calcium-fluorescence signal in response to the 1s, 100Hz train of action potentials (black bar). (c) Prosthetic hand's middle finger flexes and extends under control of the optical signal from panel b. Virtual red dot denotes the tip of the middle finger driven in the experiment. (d) Corresponding finger joint angle illustrates digit flexion occurring during supra-threshold optical control signal.*

RESULTS

An axon which fluoresced in response to the simulated motor command was selected. The calcium response originated at the center of the selected node-of-Ranvier and propagated bi-directionally into the internodal region of the axon. The nodal region, which was used for the motor command signal, showed approximately

12% change (dF/F) in fluorescence intensity. This signal amplitude was comparable to that achieved in prior work for an action potential pulse train frequency of 100 Hz [12]. Since an open-loop velocity control paradigm was employed, the digit was driven in flexion for the duration of the supra-threshold optical signal at a rate proportional to the signal intensity (about 1.5 rads/sec). Upon cessation of the command signal the finger is driven in extension at max speed (2 rads/sec) until the hand is fully open, per the cookie-crusher paradigm (**Figure 6).**

Proportional Control was demonstrated using previously recorded signals collected for a range of action potential pulse trains frequencies which were then used post-hoc to drive fingers in a position-control paradigm. As characterized in earlier work **[10]** average fluorescence amplitudes of sustained stimulus are linearly modulated by the action potential pulse train frequency. Such graded signals therefore encode intensity of the motor command. The fingers flexed to a position proportional to the intensity change produced by action potential pulse train frequency which was modulated between 25-125Hz (**Figure 7).**

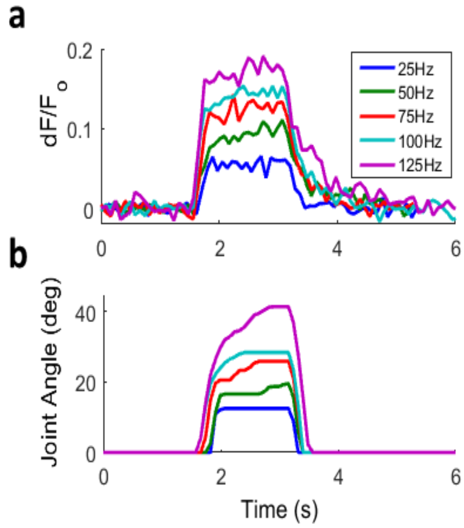

Figure 7: *Motor flexion of the prosthetic digit is graded by the action potential pulse train frequency of the optical calcium signal. (a) Graded calcium-fluorescence transients in an axon node-of-Ranvier in response to a range of action potential frequencies. (b) Resulting finger joint angles of the prosthetic finger as driven with control signals from panel a.*

While previous studies have optically stimulated peripheral nerve axons for functional modulation of motor units [13-15] using the light-activated ChannelRhodopsin2 (ChR2) there is an absence of literature describing the use of optically obtained signals from peripheral axon activity for device control. However, the range of action potential frequencies used to drive the prosthesis in this study is within a physiologically relevant range since action potential pulse train frequency typically varies between 15-500Hz (in the non-refractory range). The control signal was derived from a 1 second, 100 Hz action potential burst would likely correspond to a low-side motor command. The present experiments demonstrate the potential for read-out and control using an *ex vivo* model. In other work [11] we have demonstrated that similar signals (dF/F) can be obtained using a genetically encoded calcium indicator, GCaMP, with a retro-viral (rAAV) delivery.

CONCLUSIONS

Proof-of-concept for an optogenetic nerve interface is demonstrated by showing our ability to read-out signals at the axonal scale in real-time and apply it to the control of a prosthetic hand. Optical signals generated by frequency modulated action potentials in an axon were transduced to provide proportional prosthetic finger actuation.

ACKNOWLEDGEMENTS

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THE MYOKINETIC CONTROL INTERFACE: HOW MANY MAGNETS CAN BE IMPLANTED IN AN AMPUTATED FOREARM? EVIDENCES FROM A SIMULATED ENVIRONMENT

Marta Gherardini[a,b], Federico Masiero[a,b], Stefano Milici[a,b], Francesco Clemente[a,b], Christian Cipriani[a,b]

[a] *The Biorobotics Institute, Scuola Superiore Sant'Anna, 56127 Pisa, Italy*
[b] *Department of Excellence in Robotics & AI, Scuola Superiore Sant'Anna, 56127 Pisa, Italy*

ABSTRACT

The displacement of residual muscles during voluntary contraction in a transradial amputee could be effectively exploited to control multiple degrees of freedom in a hand prosthesis. We recently introduced a new human-machine interface (the *myokinetic control interface*) which aims at tracking muscles contraction through implanted permanent magnets and magnetic field sensors located inside the socket. Magnetic markers (MM) tracking systems have been widely investigated in the past, especially for controlling and guiding medical tools for intra-body applications. However, specific design rules for a multiarticulate robotic hand control system have not been defined yet. Here, we studied the tracking accuracy of multiple implanted magnets by simulating different levels of trans-radial amputation using a 3D CAD model of the forearm. A magnets placing procedure was developed to position the MMs in the available muscles, following general guidelines derived in our previous study. The localizer was able to accurately track up to 9, 13 and 18 MMs, in a proximal, middle and distal representative amputation, respectively. Localization errors below ~3% the length of the trajectories travelled by the MMs during muscles contraction were achieved for all amputation levels. Not only this work answers the question: "how many magnets could be implanted in a forearm and successfully tracked with a myokinetic control approach?", but also provides interesting insights for a wide range of bioengineering applications exploiting remote tracking.

INTRODUCTION

An upper extremity amputation is an event that profoundly affects the quality of life in several aspects, limiting the individual in performing working and daily living activities. Commercially available artificial hands and arms are often controlled through surface EMG electrodes that record the electrical activity generated by the residual muscles when contracting. Often, this approach suffers the lack of accessible independent control sources, and its performance is thus limited in the case of multi-articulated prostheses [1]. In the last years, different solutions have been proposed to overcome this limitation and increase the number of degrees of freedom (DoFs) that can be controlled independently.

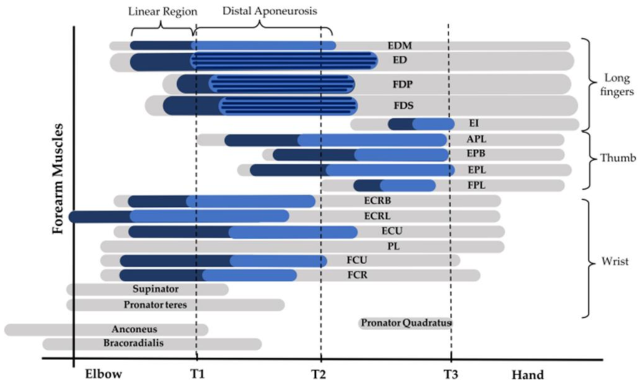

Figure 1: Detail on the anatomical distribution of the extrinsic muscles of the hand inside the forearm. The colored sections indicate the regions of the muscles belly which were considered for the implant. Dark blue regions indicate the proximal sections, which contract proportionally to their distance from the muscle origin. Light blue regions indicate the distal sections, which move by an amount corresponding to the maximum physical contraction. Gray muscles/sections were instead excluded. The three simulated levels of amputation (T1, T2, T3) are also indicated.

New technologies like wireless implantable myoelectric sensors (IMES) [2] or epimysial electrodes wired through osseointegrated implants [3], enabled direct interfacing with the physiological structures involved in the motor control, when still intact. Our group recently proposed a new concept of human-machine interface for the control of artificial limbs that takes advantage of magnetic tracking, termed *myokinetic control interface* [4]. The idea is to implant multiple permanent magnets (magnetic markers – MMs) into the residual muscles of an amputee, track their movements using magnetic sensors hosted in the socket, and use these signals as control inputs in a prosthesis, e.g. a hand. Notably, localizing the implanted magnets is equivalent to measure the contraction/elongation of the muscle they are implanted in, as the magnets move with it.

Most of the magnetic tracking systems proposed so far reconstruct the pose of a single marker using an appropriate number of sensors. Exceptions to single marker systems are the trackers developed by Yang et al. [5], Taylor at al. [6] and Tarantino et al. [7], that considered the pose of three (15 unknowns), four (20 unknowns), and seven (35 unknowns) markers. A more recent study suggested that this limit could be overcome and that, theoretically, an indefinite high number of magnets could be properly tracked, as long as some general design rules are respected [8]. Here, we sought to transfer such findings into a more realistic scenario, and to validate them by simulating the tracking of multiple magnets implanted in an anatomically appropriate model.

Thus, using a 3D CAD anatomical model of the forearm, we simulated the implant of n magnets in n available independent muscles. Three representative amputation levels were studied, in which both n and the implant sites were defined based on the forearm anatomy and on the rules identified in our previous study [8]. Results showed that it was possible to track up to 9, 13 or 18 MMs in a proximal, middle or distal representative transradial amputations, respectively. Remarkably accurate tracking performance were achieved, as localization errors always proved below ~3% the entire trajectories of the MMs inside each muscle. These outcomes are relevant because they suggest that a large number of magnets could be implanted and effectively tracked, thus allowing to achieve independent control of multiple DoFs in a hand prosthesis.

MATERIALS AND METHODS

Three configurations resembling three possible levels of transradial amputation were simulated with the aid of a 3D CAD model of the forearm of a healthy human (50th percentile male; Zygote, American Fork, US). The first and second configurations (T1 and T2) simulated amputations occurring at the first and second proximal third of the forearm, respectively (Fig. 1). The third one accounted for an amputation across the carpal bones (wrist disarticulation) leaving most of the extrinsic hand muscles available for the implant (T3, Fig. 1). For each configuration, we first identified n, i.e. the number of magnets that could be implanted and independently tracked, and defined their position in the muscles. This was done through a placing procedure that took into account: (i) the geometry of the residual forearm; (ii) a simplified biomechanical model of the muscle contraction; (iii) general guidelines identified in our previous study [8]. We then simulated the MMs movement caused by the muscles contraction, and acquired the generated magnetic field through N simulated sensors. Finally, we ran a localization algorithm to estimate the MMs poses to verify the effectiveness of the placement procedure.

MMs were modelled as Nd-Fe-B N45 grade cylindrical magnets (axial remanent magnetization $B_r = 1.27$ T, radius = 1 mm, height = 2 mm). Only muscles that after the amputation had a residual length of at least 20% the original one [9] were considered eligible for the implant. According to such criterion, T1, T2 and T3 presented a total of 18, 19 and 23 eligible muscles, respectively (Fig. 1). The displacement of the muscles was modelled according to the following linear equation:

$$d(x) = \begin{cases} \frac{x}{L} d_{max}, & x < L \\ d_{max}, & x \geq L \end{cases} \qquad (1)$$

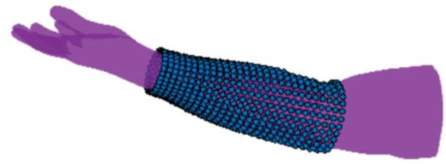

Figure 2: Sensors arrangement around the forearm, extracted from the CAD model. A grid of N = 840 sensors was used to collect the magnetic field generated by the implanted magnets. The longitudinal and radial step of the grid was set to 10 mm and 12°, respectively.

where x is the coordinate identifying the curve that runs along the muscle central axis, starting at the ideal transition between the proximal tendon and its belly; $d(x)$ indicates the actual muscle displacement; d_{max} is the maximum muscle contraction (assumed equal to 10 mm); L is the length of the muscle belly at rest.

Magnets Placing Procedure

A magnets placing procedure was implemented for defining their initial (rest) position in the residual muscles for each configuration. More specifically, we exploited an optimization procedure based on a non-linear programming solver implemented in Matlab (MathWorks, Natick, MA). Such procedure worked under the following hypotheses: (i) only one magnet could be implanted in each muscle; (ii) muscles deformation was assumed to take place only in the longitudinal direction; (iii) magnets could be implanted only in (sections of) the muscle belly, and not in the tendons; (iv) the magnetic moment vectors of the implanted MMs always pointed radially, in order to maximize the magnetic field measured by the sensors.

The procedure was initialized by placing the MMs in the center of the available muscles belly (i.e., the midpoint of the central axis). Then, it searched for a placement of the MMs that maximized both the average $d(x)$ and the average value of the geometrical parameter $\bar{R}$, defined as:

$$\bar{R} = \frac{1}{n} \sum_{i=1}^{n} \frac{L_{inter-MM_i}}{L_{MM-sensor_i}} \qquad (2)$$

where $L_{inter-MM_i}$ indicates the distance between the i-th MM and the nearest implanted MM and $L_{MM-sensor_i}$ is the distance between the i-th MM and the nearest sensor. The procedure searched for a spatial arrangement that ensured $R_i \geq 0.6$ for each magnet, a condition supposed to guarantee an accurate multi-magnet tracking [8]. Initially, a number of MMs equal to the total number of sites eligible for implantat was considered. Then, at the end of each iteration, the magnet that scored the lowest R (if below 0.6) was removed. The placement procedure was iterated until all

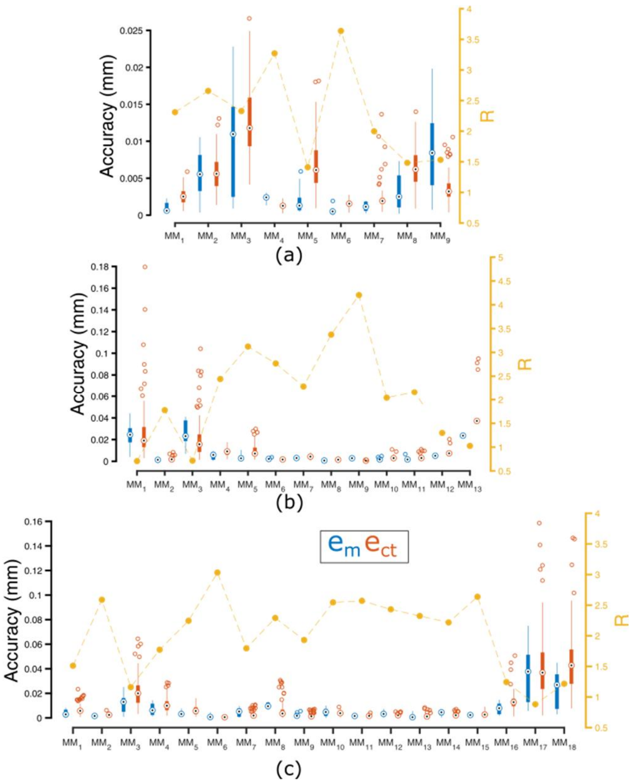

Fig. 3. Boxplots represent e_m (blue) and e_{ct} (red) for each MM, while the lines indicate the corresponding R value, for (a) T1, (b) T2 and (c) T3. In general, lower R values led to lower accuracies, and vice-versa.

remaining MMs were successfully placed (all $R_i \geq 0.6$) within the simulated forearm.

Localization Problem

Once the placing procedure was completed, the MMs were moved along the muscle axis following anatomically appropriate trajectories. Each trajectory originated in the position defined by the placing procedure, and ended proximally at a distance $d_i(x)$. The MMs displacement was approximated by translating them, one at a time, along 11 equidistant checkpoints (0%, 10%, 20%, …, and 100% the trajectory length). At each checkpoint, the analytical model described in [10] and validated in [11] was used to simulate the magnetic field generated by the MMs. Indeed, analytical approaches generally have a lower computational cost compared to numerical methods, other than being more accurate [12]. Consequently, many studies focused on the analytical calculation of the magnetic field produced by currents in a coil [13], or by arc-shaped permanent magnets (e.g. cylindrical permanent magnets) [11]. To reduce the computational burden, these analytical formulations are generally expressed in terms of complete elliptic integrals or through series expansion [14]. A compact and efficient representation of the magnetic field produced by an axially uniformly magnetized cylindrical permanent magnet was used in this work [10].

Such field was sampled on a grid of $N = 840$ simulated sensors arranged around the forearm, and ideally hosted within the prosthetic socket (Fig. 2). The longitudinal and radial step of the grid was set to 10 mm and 12°, respectively. This resulted in a distance between adjacent sensors between 6 mm and 10 mm. Sensors recordings at each checkpoint were stored and subsequently fed to a Matlab script that ran the Levenberg-Marquardt algorithm [15] to retrieve the poses of the MMs offline.

To solve the inverse problem of magnetostatics and thus retrieve the poses of the MMs, the latter were approximated as point-like dipoles, akin to our previous works [4][7][8]. The localization accuracy, both in terms of position and orientation, was evaluated as:

$$E_p \approx e_m + e_{ct} \tag{3}$$

where e_m accounts for inaccuracies in tracking the displacement of the moving magnet (i.e., model error), while e_{ct} accounts for false predictions of simultaneous displacement affecting the non-moving magnets (i.e., cross-talk effect). e_m and e_{ct} were defined as the Euclidean distance between the actual and the estimated displacement for the moving and non-moving MMs, respectively, akin to our previous works [4][7][8].

RESULTS

The placement procedure selected a total of nine target muscles for T1, 13 for T2 and 18 for T3. The minimum and maximum displacement underwent by the MMs were respectively 0.6 and 1 mm, across all configurations. For the sake of brevity we report only the results related to the position accuracy. In fact, the orientation accuracy proved always lower than 0.36° for all MMs in all configurations, and its trend closely matched that found for the position accuracy.

Regarding the localization accuracy, e_m proved always lower than 0.07 mm (Fig. 3). In particular, the highest e_m values were obtained for MM_3 in configuration T1 (e_m= 0.02 mm), for MM_{13} in configuration T2 (e_m= 0.05 mm), and MM_{17} in configuration T3 (e_m= 0.07 mm). e_{ct} proved generally higher than e_m (Fig. 3), but still in the same order of magnitude. Indeed, the highest e_{ct} values obtained were 0.03 mm for MM_3 in configuration T1, 0.18 mm for MM_1 in configuration T2, and 0.16 mm for MM_{17} in configuration T3. Overall, both e_m and e_{ct} proved lower than or equal to 3% the shortest trajectory covered by the MMs during muscle contraction (i.e., 6 mm). For both e_m and e_{ct}, the localization accuracy generally worsened when R decreased (Fig. 3). As an example, in configuration T2, MM_1 showed the highest e_{ct} (equal to 0.18; R = 0.71), while it proved always lower than 0.01 for MM_9 (R = 4.20).

DISCUSSION

In this work, we studied the effects of the complex anatomy of the human forearm on the design of a myokinetic control interface aimed at driving multiple DoFs in a hand prosthesis. A magnets placing procedure was implemented which, based on the forearm anatomy and on predefined rules, guided the choice of the number and sites of implant of multiple MMs.

Traumatic amputations lead to a large variety of muscle health conditions, which are not standardized across subjects. Our approach allows to customize the MMs arrangement based on the muscles distribution of a specific patient. The latter could be made available from 3D MRI images [16]. By enabling a preclinical planning of the MMs placement, we could significantly reduce the duration of the surgical procedure, and optimize as well as ensure good performance of the prosthesis control system.

In agreement with previous findings [7][8], localization errors generally increased for lower *R* values. This justifies the need for planning the MMs placement as opposed to a random one. Specifically, in [7] nine MMs were randomly distributed in a workspace that mimicked only the bulk volume of the forearm. In this case, the system failed to retrieve their poses with acceptable accuracy. Here, the imposed constraints allowed to achieve good tracking accuracies even when the number of magnets was doubled.

The present study was indeed limited in some respects. First, in order to limit the number of combinations tested, the orientation of the MMs was kept fixed (pointing towards the sensors). This is the optimal configuration, selected to maximize the magnetic field sampled by the sensors. However, variability in the orientation of the MMs is expected as these will likely be implanted manually in the muscles. Secondly, we considered a simplified linear model to describe the muscle displacement. This only captured longitudinal elongations/contractions of the muscles, while it is known that these undergo radial deformations as well. Thus, the ability of the approach in coping with more complex movements of the MMs remains to be tested.

The outcomes of this work pave the way towards the development of an intuitive control system that can be used to drive a dexterous hand prosthesis, by significantly improving both the naturalness of the control strategy and its functionality. Furthermore, they are of great interests for a multitude of bioengineering applications that exploit multi-magnet tracking in a constrained workspace.

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MYOKINETIC PROSTHESIS CONTROL ORIENTED ENVIRONMENTAL MAGNETIC DISTURB ANALYSIS

Valerio Ianniciello[a,b], Marta Gherardini[a,b], Francesco Clemente[a,b], Christian Cipriani[a,b]

[a] *The Biorobotics Institute, Scuola Superiore Sant'Anna, 56127 Pisa, Italy*
[b] *Department of Excellence in Robotics & AI, Scuola Superiore Sant'Anna, 56127 Pisa, Italy*

ABSTRACT

Several medical applications involve the use of remote magnet tracking for retrieving the position of tools instrumented with one or more magnets, when a free line-of-sight between the magnets and the tracker is not available. Our group recently proposed to implant passive magnetic markers (i.e. permanent magnets) in the forearm muscles of an amputee in order to track the displacements of those muscles during contraction. The idea is to use the retrieved information to control a hand prosthesis. We called this the *myokinetic control interface*. However, besides the system feasibility, how much its accuracy and precision are affected by external noise sources has not been quantified yet.

Here, through an experimental setup, we investigated the influence of different magnetic/electromagnetic interferences on the localization accuracy of three permanent magnets. The magnetic field generated by the magnets was collected both in interference-free conditions and in presence of disturbances. Localization errors achieved under different conditions, and for both raw and low-pass filtered signals, were derived. Results showed that the steel bar caused the maximum average localization error, equal to 9.8 mm and 74° in terms of position and orientation, respectively. The microwave oven caused instead the maximum localization variability, with a standard deviation of 0.21 mm and 2.2°. The low-pass filtering operation (5 Hz cut-off frequency) did not lead to significant improvement in the accuracy, resulting in an error decrease always below 7% compared to the unfiltered signals.

This work is important because it gives a quantitative measure of the disturbances encountered in everyday life which could cause the failure of those systems exploiting remote tracking.

INTRODUCTION

The deprivation of a hand is an event that significantly affects a person's ability in performing working and daily living activities (AdL), thus having a strong impact on his/her social life. In order to restore the lost motor functions in individuals with a hand amputation, two important factors are needed: first, the development of a dexterous prosthesis; secondly, the development of an intuitive Human-Machine Interface (HMI). Despite the recent research efforts to find a solution to these problems, both are still far from being solved. Indeed, on one hand a prosthesis able to replicate the dexterity of the natural limb has not been realized yet; on the other, commercially available prostheses often have more Degrees of Freedom (DoFs) than those controllable with current control strategies. In this regard, commercial hand prostheses are currently driven through HMIs which exploit the so-called direct control [1]. The latter consists in mapping the EMG signal recorded from agonist/antagonist muscle pairs using surface electrodes to a unique function in the prosthesis. Despite being intuitive and robust [2], this approach is hardly applicable to the control of multiple functions/DoFs, due to the lack of accessible independent control sources [3].

In order to overcome this limit and enhance the number of naturally controllable DoFs, our group recently introduced an alternative solution, dubbed the *myokinetic control interface* [4]. The idea is to implant permanent magnets into the residual forearm muscles, track their displacement using external magnetic sensors, and use this information as control input for the prosthesis. This approach would allow to physiologically (i.e. simultaneously and proportionally) control multiple, independent DoFs of the prosthesis by exploiting simple, passive implants.

In our previous work [5], we presented an embedded system which proved able to accurately localize in real-time up to five magnets. Such a system could potentially be used to control a robotic hand/arm. In order for this technology to be used in real-world scenarios, we need to make it robust against an environment which is largely corrupted with noise. Indeed, the presence of ferromagnetic elements and electromagnetic noise can potentially compromise the accuracy of the magnet tracking system. This problem has been poorly studied, since most of magnet tracking applications found in the literature are carried out in dedicated environments (e.g. an operating room), where the different noise sources can be avoided or modelled [6].

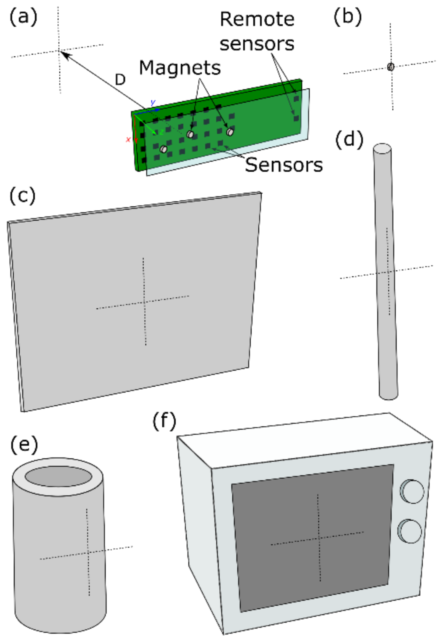

Figure 1: Experimental setup. The acquisition board and the three magnets to be localized are shown w.r.t. the disturbing object position (a). Those objects are: the magnet (b), the steel sheet (c), the steel bar (d), the hollow steel cylinder (e) and the microwave oven (f).

Preliminary results on the efficacy of a shielding strategy were presented in our previous study [4]. However, a deeper understanding of the problem in a real-world scenario remains to be tested. For this reason, it is of pivotal importance to investigate the effects of different interferences in a real experimental setup, in order to quantify the entity of the disturbances and find solutions to ultimately reject them.

MATERIALS AND METHODS

System Architecture

A *myokinetic control interface* is composed by two elements, namely the implanted magnets and a localizer embedded into the prosthetic socket which is responsible for continuously retrieving their poses (localization process). The latter is achieved by solving the so-called inverse problem of magnetostatics which, akin to previous works [4][5][7], was done by exploiting the well-known Levenberg–Marquardt optimization algorithm [8]. The acquisition unit already introduced in our previous work [5] was used to collect the magnetic field generated by the magnets. It consists of a custom board that collects signals from 32 three-axis magnetic field sensors (MAG3110, NXP Semiconductors NV, Eindhoven, Netherlands; full-scale output of ±10 G and sensitivity of 1 mG), arranged in a 4 × 8 matrix, except for two sensors that are placed remotely to compensate for the geomagnetic field. The sensors are connected to a 16-bit architecture microcontroller (dsPIC33EP512MU810-I/PT, Microchip Technology Inc., Chandler, AZ, USA) which samples their readings and transmits them to the actual localizer.

In our previous work [5] we proved the equivalence in terms of localization precision and accuracy between the embedded localizer implemented in C (running on a MIMXRT1050–EVKB, NXP Semiconductors, Eindhoven, NL) and the PC implementation of the same algorithm in Matlab (MathWorks, Natick, MA, USA). In this work, the latter was used in order to simplify the data analysis phase.

Experimental Setup

There exist different error sources which can affect the magnets pose estimation. Those due to model approximations, cross-talk effect between magnets, as well as sensor fluctuations have been extensively studied in our previous works [4][5][7]. Here, we addressed localization inaccuracies caused by environmental factors through a dedicated experimental setup (Figure 1).

Specifically, three axially magnetized neodymium cylindrical magnets (d = 4 mm; h = 2 mm; M = 0.0254 A·m^2 and B_r = 1.27 T) were fixed in anatomically relevant positions w.r.t the acquisition unit, using a rigid frame. Their magnetic field in presence of different noise sources (Table 1) was acquired and stored for offline analysis. Such interferences included the presence of close magnetic/ferromagnetic objects, as well as active electromagnetic noise sources (e.g. microwave oven, moving elevator).

Table 1: Settings details

	Disturbance	Distance (D)	Notes
Representative elements	Magnet	5 cm; 15 cm; 25 cm.	Same type of magnet as those used for the localization process.
	Steel bar		C40 steel. D = 2.5 cm, h = 80 cm
	Hollow steel cylinder		C40 steel. D_{ext} = 1.3 cm, d_{int} = 67 cm, h = 24 cm
	Steel sheet		0.5 mm thick stainless steel. l_1 = 32 cm, l_2 = 27 cm
AdL elements	Microwave oven		Samsung, model GW712K. Acquired with power set to 750 W.
	Electrical substation	40 cm	Distance is from the substation door.
	Elevator	70 cm	Distance is from the floor. Produced by BAMA srl. Acquired both while elevator is still and moving.
Ref.	Disturbance-free	-	Acquisition unit held still with no disturbance. Used as reference.

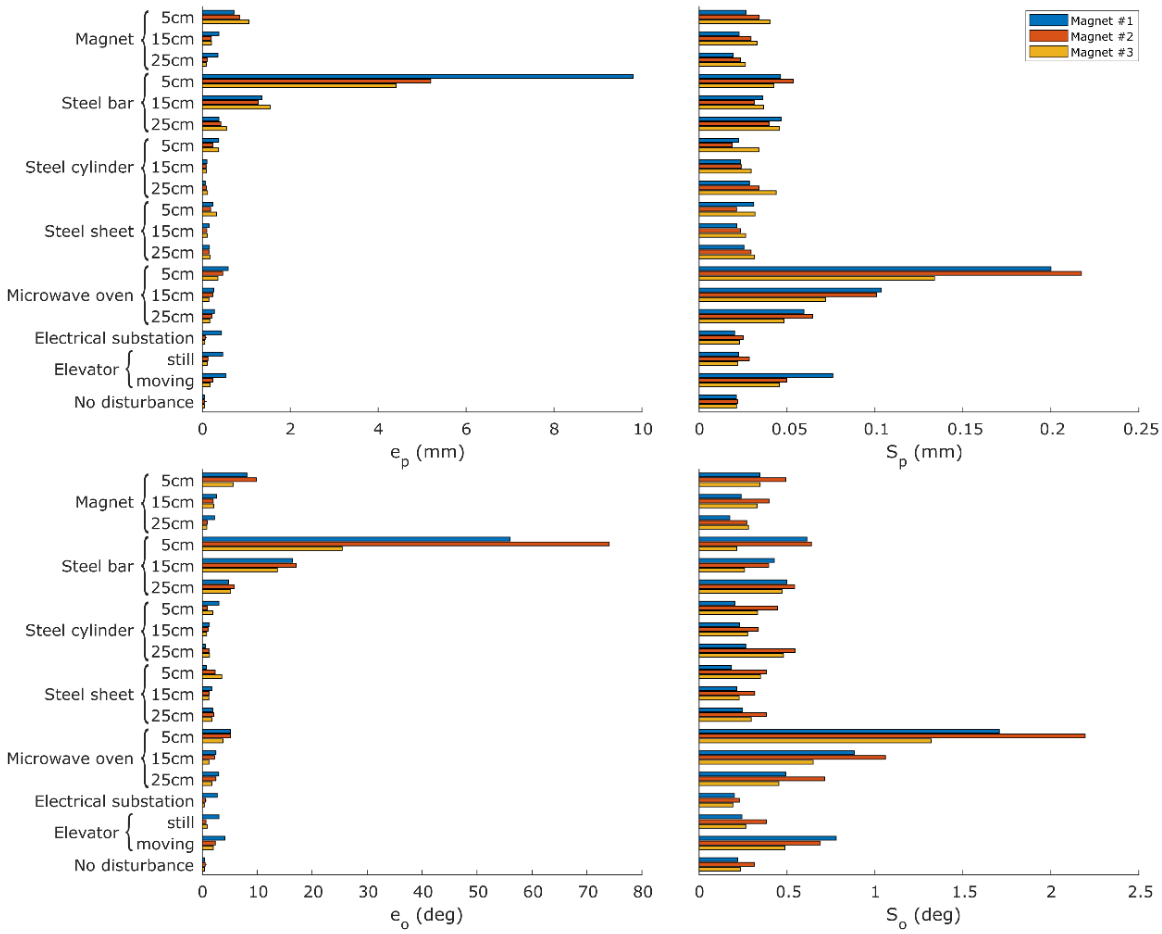

Figure 2: Results. Average position and orientation errors (e_p, e_o) and relative standard deviations (S_p, S_o) for the three magnets in all the experimented settings.

The tested noise sources were selected as representative of objects that can be encountered in AdL. For instance, a locker gives an interference comparable to that of a metal sheet, while a mobile phone originates a disturbance due to the two magnets present in its speakers. Some noise signals were acquired by considering different board - noise source distances (namely, 5 cm, 15 cm and 25 cm), because it was interesting to see how they affected the localization process for different interference intensity. Others, instead, were only measured at a single distance (Table 1). 2000 samples per configuration were acquired with a sampling frequency of 13 Hz, resulting in ~150 seconds per recording session. A pre-processing step was implemented by subtracting the field measured by the remote sensors and subsequently applying a low pass filter with a 5 Hz cut-off frequency, which we considered a reasonable bandwidth for human movements. Both the raw and the low-pass filtered acquisitions were used for estimating the pose (i.e. position and orientation) of the three magnets, in order to compare the results.

For assessing the entity of the disturbance, the average position and orientation errors (e_p, e_o) and their relative standard deviations (S_p, S_o) were derived. In order to isolate the noise contribution from the model and the cross-talk error, the mean value of the magnets poses derived using the interference-free signals were considered as a ground-truth reference.

RESULTS

The average localization error in terms of both position and orientation proved generally higher in presence of the interference cause by the steel bar (Figure 2). In particular, a maximum e_p of 9.8 mm was shown by magnet #1 when the steel bar was put at the minimum tested distance (i.e., 5 cm). Such error was comparable to the expected range of motion of the implanted magnets inside the muscle, which is ~10 mm [9]. In the same configuration, magnet #2 showed the maximum e_o across all configurations, equal to 74°. All other noise sources generally led to a lower accuracy deterioration, resulting in a maximum e_p (e_o) value of 1.1 mm (9.8°) in presence of the magnet at 5 cm.

The variabilities S_p and S_o acquired in presence of active noise sources proved significantly higher than those derived from the disturbance-free acquisitions (Figure 2). In particular, the maximum variability was always caused by the presence of the microwave, for which a maximum S_p and S_o value of 0.21 mm and 2.2° were derived, respectively. Ferromagnetic noise sources showed instead a smaller variability.

The 5 Hz low-pass filtering operation generally led to a small error reduction when compared to the error obtained using the raw signals (<7% reduction). Indeed, a median relative error reduction of less than 1% for both e_p and e_o, and between 5% and 7% for S_p was derived (Figure 3).

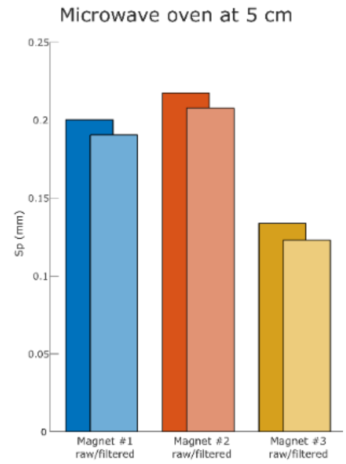

Figure 3: Comparison between the position error standard deviation (S_p) in localizations from raw and filtered data in the microwave oven at 5 cm.

DISCUSSION

In this work, we evaluated the effect of different noise sources that can be encountered in everyday life and can affect the tracking performance of a *myokinetic control interface*. Everyday objects and dedicated tools that mimic their characteristics were studied to assess this effect. Our results showed that, while it is possible to retrieve the poses of the three magnets under different noisy conditions, some settings led to a significant deterioration of the localization accuracy.

Indeed, we found that specific ferromagnetic materials caused a considerable shift in the pose estimation, comparable to what is expected to be the range of motion of the magnets inside the muscles. Furthermore, some active noise sources (e.g. the microwave oven) induced high frequency oscillations in the estimated poses. The latter could be removed by applying a proper filtering approach. The cut-off frequency used in this work (5 Hz) led to modest improvements in the localization results, and we hypothesize this to be due to aliasing effects. Indeed, since the acquisition unit worked at 13 Hz, while the active noise sources may present higher frequencies, we were not able to entirely remove the interferences. However, by setting a higher frequency for the sensors acquisition, we can expect the noise contribution to be better discriminated and filtered. This remains to be tested in future works. Nevertheless, in most cases, the entity of the localization variability (S_p and S_o) was not relevant compared to the expected magnets range of motion (Figure 2).

The outcomes of this work demonstrate the importance of developing a magnetic shielding system able to make the *myokinetic control interface* robust and reliable in a real-world scenario. They are important because they provide a quantitative measure of the disturbances that could cause the failure of remote tracking applications.

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A NOVEL METHOD FOR ACHIEVING DEXTEROUS, PROPORTIONAL PROSTHETIC CONTROL USING SONOMYOGRAPHY

Susannah Engdahl[1], Biswarup Mukherjee[1,2], Nima Akhlaghi[1], Ananya Dhawan[1], Ahmed Bashatah[1], Shriniwas Patwardhan[1], Rahsaan Holley[3], Rahul Kaliki[4], Brian Monroe[5], and Siddhartha Sikdar[1,2]

[1]*Department of Bioengineering, George Mason University, Fairfax, VA*
[2]*Center for Adaptive Systems of Brain-Body Interactions, Fairfax, VA*
[3]*MedStar National Rehabilitation Hospital, Washington, DC*
[4]*Infinite Biomedical Technologies, Baltimore, MD*
[5]*Hanger Clinic, Laurel, MD*

ABSTRACT

Although myoelectric upper limb prostheses have been commercially available for decades, many patients who receive these devices abandon them due to their limited functionality. Some of these functional limitations are related to the difficulties in sensing activity in different muscle compartments with surface electromyography. We believe it is possible to overcome the limitations of myoelectric control through use of sonomyography, or ultrasound-based sensing of muscle deformation. Sonomyography can better distinguish individual muscle activity and provides access to control signals that are directly proportional to muscle deformation, which has the potential to significantly improve prosthesis functionality. In this paper, we will describe our work towards developing a low-power wearable imaging system that will enable sonomyographically-controlled prostheses.

INTRODUCTION

Major upper limb loss affects more than 41,000 individuals in the United States alone [1] and can cause significant functional deficits. Despite recent advances in electromechanical design for prosthetic hands, development of control strategies has not kept pace. Surface electromyography (EMG) remains the predominant method for sensing muscle activity in order to actuate a prosthetic hand. As a result of the poor amplitude resolution and low signal-to-noise ratio [2], [3] for EMG signals, it can be challenging for prosthesis users to control a hand having more than one degree of freedom (DoF). More sophisticated signal processing algorithms relying on pattern recognition (e.g., [4]) can enable control of multiple DoFs but do not avoid the inherent problem of poor amplitude resolution. Alternative strategies such as targeted muscle reinnervation [5] or implanted electrodes [6] offer access to a richer set of control signals at the expense of surgical intervention and may not be tolerated by all individuals [7].

To address these problems, we propose the use of a different sensing modality based on ultrasound. Sonomyography (SMG), or ultrasound-based imaging of muscle contractions, is a non-invasive technique that can spatially resolve individual muscles with sub-millimeter precision. Other research groups (e.g., [8], [9]) have demonstrated that SMG is a viable option for classifying individual finger motion. We believe our group is the first to develop a low-power wearable imaging system for SMG that ultimately could be incorporated into a prosthesis socket, and demonstrate the ability of SMG to enable proportional positional control in amputee subjects. In this paper, we will describe our prior work in this area, present some new previously-unpublished results, as well as describe the anticipated directions for our future work.

BASICS OF SONOMYOGRAPHY

Sonomyography involves real-time ultrasound imaging of muscle contractions during voluntary movement. For different movements that the individual performs, a unique group of muscle compartments are activated and undergo mechanical deformation. Ultrasound images capture a spatially- and temporally-resolved view of the forearm muscle deformation over time as the users perform different motions (Figure 1A). Individual ultrasound scanlines can be visualized over time as M-mode images, which can be used to track deformation of specific muscle compartments (Figure 1B). By applying custom-developed image analysis and machine learning methods to these images, we can extract control signals and use them to drive a prosthetic hand (Figure 1C).

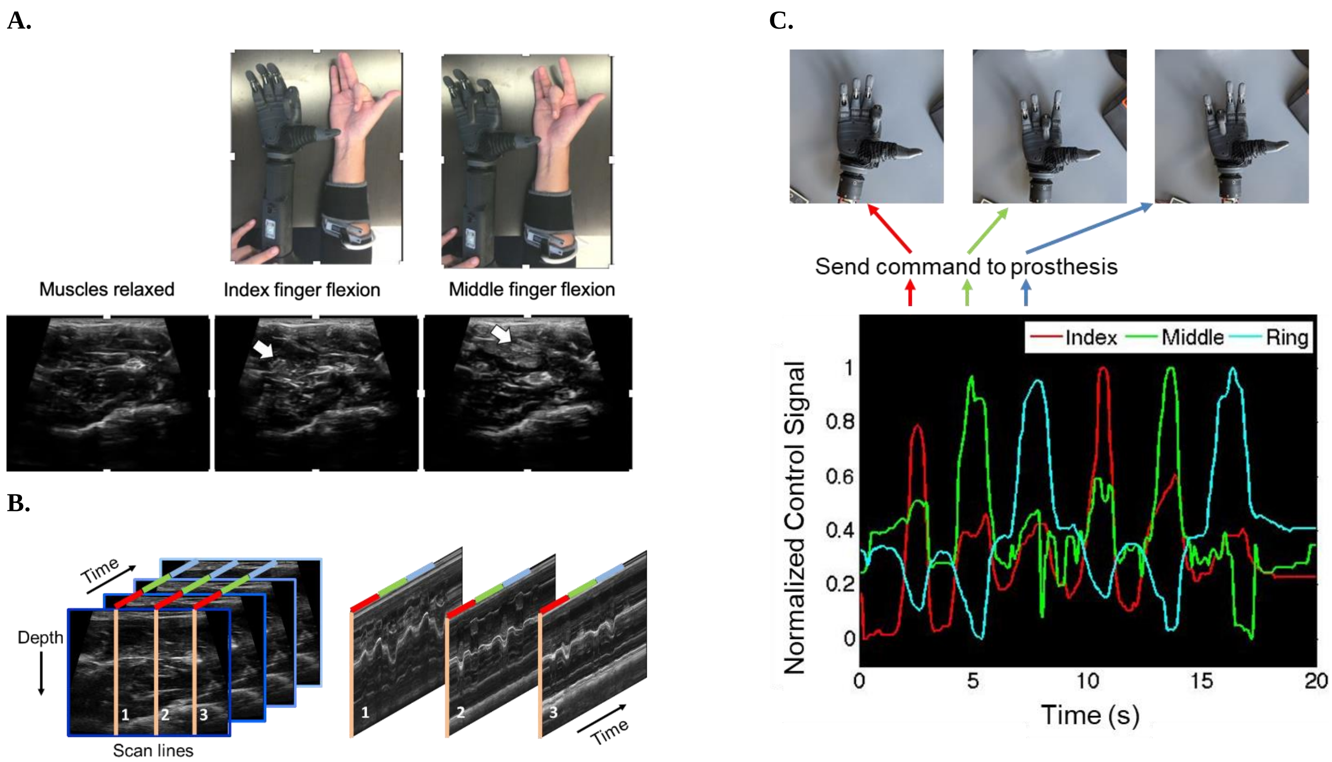

Figure 1. Schematic showing our approach to prosthetic control with SMG. (A) Muscle deformation over time is tracked with an ultrasound transducer placed on the forearm. The figure shows an able-bodied subject performing index finger flexion and middle finger flexion. The corresponding ultrasound images show different muscle compartments deforming for each movement. (B) M-mode images (depth over time) show deformation of different muscle compartments over time corresponding to individual finger movements (red, green, blue segments). (C) Control signals are extracted based on the muscle deformation associated with individual finger movements (red, green, blue traces) and are then mapped to movement of a prosthetic hand.

GRASP CLASSIFICATION USING SONOMYOGRAPHY

Ultrasound systems are becoming increasingly inexpensive and portable, and probes can now be connected to a laptop or smartphone through USB (e.g., Philips Lumify, Butterfly iQ, Mobisante). To demonstrate the feasibility of using these systems as part of a wearable prosthesis system, in a previous study, we recruited 10 able-bodied individuals and attached a handheld ultrasound probe to their forearm. Participants were asked to move each individual digit (thumb, index, middle, ring, little fingers) multiple times in sync with an auditory cue from a metronome. Activity pattern images corresponding to each movement were saved in a training database and were used to train a k-NN algorithm. Leave-one-out cross-validation revealed the offline classification accuracy to be 98.33%. We also demonstrated that ultrasound echogenicity changes proportionally in response to different levels of thumb flexion, showing the potential for achieving proportional digit control. Taken together, these results show that inexpensive handheld ultrasound probes are adequate for sensing and classifying muscle activity [10].

We have also explored the use of low-resolution binary activity patterns as features for classifying complex grasping gestures [11]. In a group of six able-bodied individuals, we demonstrated an average offline classification accuracy of 91% for 15 different grasps. Additionally, we showed that simultaneous wrist and hand movements (e.g., power grasp with wrist pronation) can be classified with > 90% accuracy. These results indicate that low-resolution imaging can be a viable option in a wearable ultrasound system.

As a step towards further reducing the instrumentation footprint, we investigated the effect of using a sparse set of ultrasound scanlines for gesture classification [12]. We recorded ultrasound images from the forearms of five able-bodied subjects performing five grasps (power grasp, pinch, index point, key grasp, wrist pronation) using a 128-element linear array transducer. We then selected different subsets of scanlines to quantify the extent to which classification accuracy was affected. Even with a subset of only four scanlines, classification accuracy was virtually unchanged ($94 \pm 6\%$ for 128 scanlines, $94 \pm 5\%$ for 4 scanlines). This demonstrates the feasibility of using a small number of single-element transducers rather than a full array, which

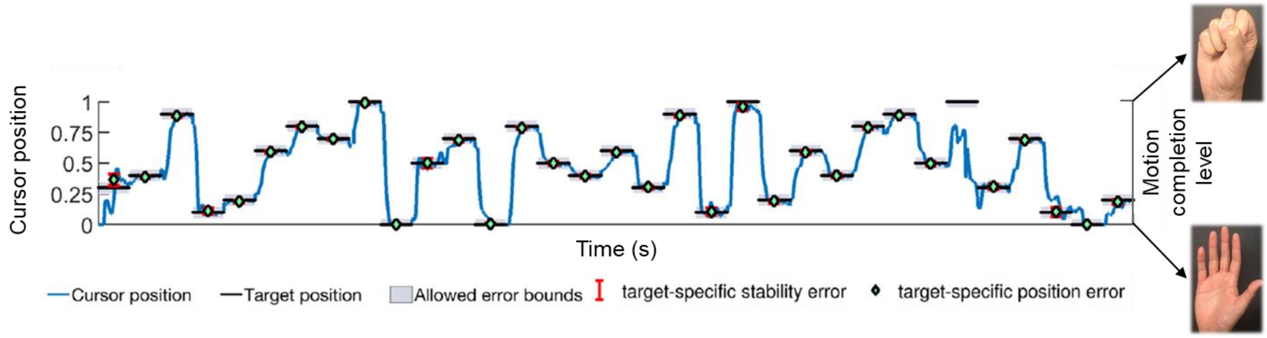

Figure 2. Representative time series plot showing an amputee performing the target acquisition task using power grasp. A cursor position of "0" corresponds to a fully relaxed state and a cursor position of "1" corresponds to motion completion.

simplifies the instrumentation that would need to be incorporated into a prosthesis socket.

SONOMYOGRAPHIC CONTROL IN AMPUTEES

Musculoskeletal anatomy can differ significantly between the forearm of an able-bodied individual and the residual limb of an individual with transradial limb loss. It is therefore important to demonstrate that our methods for classifying finger movements and complex grasps in able-bodied individuals are applicable to amputees as well.

We tested the ability of our system to distinguish between five different hand motions (power grasp, wrist pronation, tripod, key grasp, and point) in a group of five able-bodied controls and five transradial amputees. Average cross-validation accuracy was 100% for able-bodied controls and 96% for amputees, indicating that the system could reliably predict motions in both groups [13].

Having demonstrated that real-time classification of motion endpoints is possible in amputees, we next sought to implement an extended version of our algorithms that would enable proportional position control [13]. Participants were asked to perform a target acquisition task in which they manipulated a computer cursor that moved vertically in response to the degree of grasp completion (Figure 2). A series of targets were presented at 10 different levels of grasp completion and we quantified participants' ability to reach each target and stay within the target bounds. The task was repeated for each of the five hand motions. There were no differences in performance between groups, showing that sonomyography can enable proportional position control for both amputees and able-bodied individuals.

WEARABLE LOW-POWER ULTRASOUND SYSTEM FOR PROSTHETIC CONTROL

In our most recent work, we have developed a low-power ultrasound imaging system using a novel signalling method that uses low-voltage (5V peak to peak) transmissions. The system consists of a wearable band of 4 single element transducers (Figure 3) weighing less than 2 ounces, and benchtop instrumentation powered by a 7.4 V battery. This system consumes 350 mW/channel and provides comparable results to conventional pulse echo imaging and is well below FDA guidelines for acoustic exposure.

In a preliminary study (previously unpublished), we tested the performance of this system. All study procedures were approved by the George Mason University Institutional Review Board, and we obtained an abbreviated Investigational Device Exemption to test this system on human subjects. We recruited 5 able-bodied subjects, who were asked to perform 9 different motions: key grasp, pinch and power grasp in three different wrist orientations: supine, neutral and prone. The acquired data from the 4-channel system were then analysed offline to calculate the confusion matrix and classification accuracy. Our results show that the system can differentiate between 9 movements with 94.7% classification accuracy on average. The key grasp in supine position was the motion with the lowest classification accuracy overall.

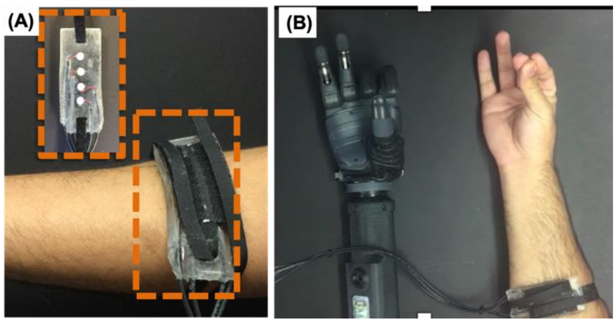

Figure 3. Wearable low-power 4-channel ultrasound system (A) for controlling a prosthetic hand (B).

DISCUSSION

We believe our research thus far demonstrates numerous advantages of SMG compared to EMG, making it a promising modality for restoring dexterous movement to individuals using upper limb prostheses. One of the primary benefits of SMG is that muscle activity can be sensed with high spatial specificity, even in deep-seated muscle compartments. As a result, cross-talk from muscles that are not associated with the intended movement is effectively suppressed. We have shown that SMG can classify five

individual digit movements with 97% accuracy [10] and 15 complex grasps with 91% accuracy [11] in able-bodied individuals. Importantly, similarly high classification rates can also be achieved in transradial amputees [13]. It is also noteworthy that full-resolution ultrasound imaging is not required to achieve these outcomes. Classification accuracies are not affected even when a subset of only four ultrasound scanlines are used. Single-element transducers may be used instead of a full array, reducing the instrumentation required for implementing SMG control in a standalone prostheses.

Furthermore, the control signals derived from muscle activity using SMG have high signal to noise and are able to resolve sub-millimeter muscle deformations, so the resultant control signals enable finely-graded proportional positional control. We have demonstrated that individuals with transradial amputation can consistently achieve 10 different graded positions using SMG [13].

Unlike the control signals derived from EMG that must be mapped to velocity of a terminal device, the control signals from SMG can be mapped to position. This strategy is similar to natural motor control and doesn't involve learning a new strategy, which is required for conventional myoelectric control. We have shown that learning to use SMG requires minimal training. In fact, transradial amputees were able to achieve 96% classification accuracy for 5 grasps after only a few minutes of training time [13].

CONTINUING WORK

The majority of our work to date has been implemented in a benchtop setting. As a next step toward demonstrating the utility of SMG control, we are now working to translate our technology to a standalone research-grade prototype that can be integrated into a prosthetic socket. This will ultimately enable functional assessment in a laboratory setting.

There are several critical questions that still need to be addressed as part of our continuing work. First, it is necessary to demonstrate whether adequate signal quality can still be maintained when the transducers are integrated into a socket and used for long periods of time. We have shown that classification accuracy is robust to changes in arm and wrist position for able-bodied individuals [11], but it remains to be seen whether coupling between the residual limb and transducer will be affected by changes in arm position, sweating, or loading introduced by the socket and terminal device. In future studies, we plan to investigate this systematically. Additionally, we will explore the extent to which use of SMG contributes to functional improvements compared to EMG. In particular, we will test whether higher scores on standard clinical tests, improved quality of movement, greater patient-reported satisfaction, and reduced cognitive load can be achieved through use of SMG.

Despite these remaining questions, we believe our work demonstrates the feasibility of using SMG to achieve real-time proportional positional control with limited training. Importantly, these outcomes can be achieve using a low-power wearable imaging system for SMG that can incorporated into a prosthesis socket. We anticipate that this approach will ultimately enable intuitive proportional control of multi-articulated prosthetic hands and will contribute to improved acceptance of these devices.

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PHANTOM HAND ACTIVATION DURING PHYSICAL TOUCH AND TARGETED TRANSCUTANEOUS ELECTRICAL NERVE STIMULATION

Luke E. Osborn[1], Courtney W. Moran[2], Erin E. Sutton[1], Adam C. G. Crego[2], Jonathan A. Forsberg[3,4], Matthew S. Fifer[1], Robert S. Armiger[1]

[1]*Research & Exploratory Development Department, Johns Hopkins University Applied Physics Laboratory, Laurel, MD, USA.*

[2]*Asymmetric Operations, Johns Hopkins University Applied Physics Laboratory, Laurel, MD, USA.*

[3]*USU-Walter Reed Department of Surgery, Walter Reed National Military Medical Center, Bethesda, MD, USA.*

[4] *Department of Orthopaedic Surgery, Johns Hopkins School of Medicine, Baltimore, MD, USA.*

ABSTRACT

Restoring the sense of touch is a critical component for a closed-loop prosthetic limb. In an upper limb amputee, we explored regions on the residual limb that elicited sensory activation of the phantom hand through either physical touch or targeted transcutaneous electrical nerve stimulation (tTENS). We found that sensory sites on the residual limb responded to either physical touch or tTENS, but typically not both. Further, some regions of the phantom hand were only activated with one of the stimulation modalities, such as the thumb or wrist. Interestingly, some locations on the phantom hand could be activated with either physical touch or tTENS but at different locations on the residual limb. Our work helps highlight potential differences in perceived location of sensory feedback depending on the stimulation modality.

INTRODUCTION

Direct neural interfaces, such as the flat interface nerve electrode (FINE) [1], and advanced surgical techniques, such as targeted muscle reinnervation (TMR) [2] and targeted sensory reinnervation (TSR) [3], [4], have enabled significant advances in providing sensory feedback to upper limb amputees. The sense of touch can be restored to the phantom hand of using direct electrical nerve stimulation [5]. Recently, researchers used bioinspired stimulation models to convey perception of texture [6], mechanical pain [7], and increase naturalness of restored tactile sensations for improved functionality [8]. Restored sensation to the phantom hand can be achieved through noninvasive approaches including cutaneous vibration [3] and targeted transcutaneous electrical nerve stimulation (tTENS) [7], [9], [10]. Recently, researchers showed that you can even elicit illusory perception of phantom hand movement during cutaneous vibration after TMR [11]. In addition to continued research on prosthesis technology, such as advanced myoelectric control methods [12], [13] and tactile sensing [7], [14], [15], work on sensory feedback is progressing quickly.

We explored the regions of phantom hand activation in an amputee using both physical touch and TENS. The purpose of the sensory mapping was to identify the similarities and differences between the two sensory activation modalities. Because sensory feedback is possible through both physical (cutaneous vibration) and electrical (TENS or direct nerve stimulation) modalities, it is important to understand the differences to provide useful and meaningful sensory information to prosthesis users.

METHODS

As a case study for comparing tactile feedback modalities, the participant discussed was a 64 year old male with a left transhumeral amputee who previously underwent TMR surgery and has an osseointegrated interface for prosthesis attachment in his residual limb. The participant provided written informed consent to be a part of this study. This research protocol was reviewed and approved by the Johns Hopkins Medicine Institutional Review Boards in accordance with all applicable Federal regulations governing the protection of humans in research.

Sensory stimulation of the participant's phantom hand was achieved through either physical touch or targeted TENS. To active the phantom hand with physical touch, the participant used his intact hand to identify and palpate known

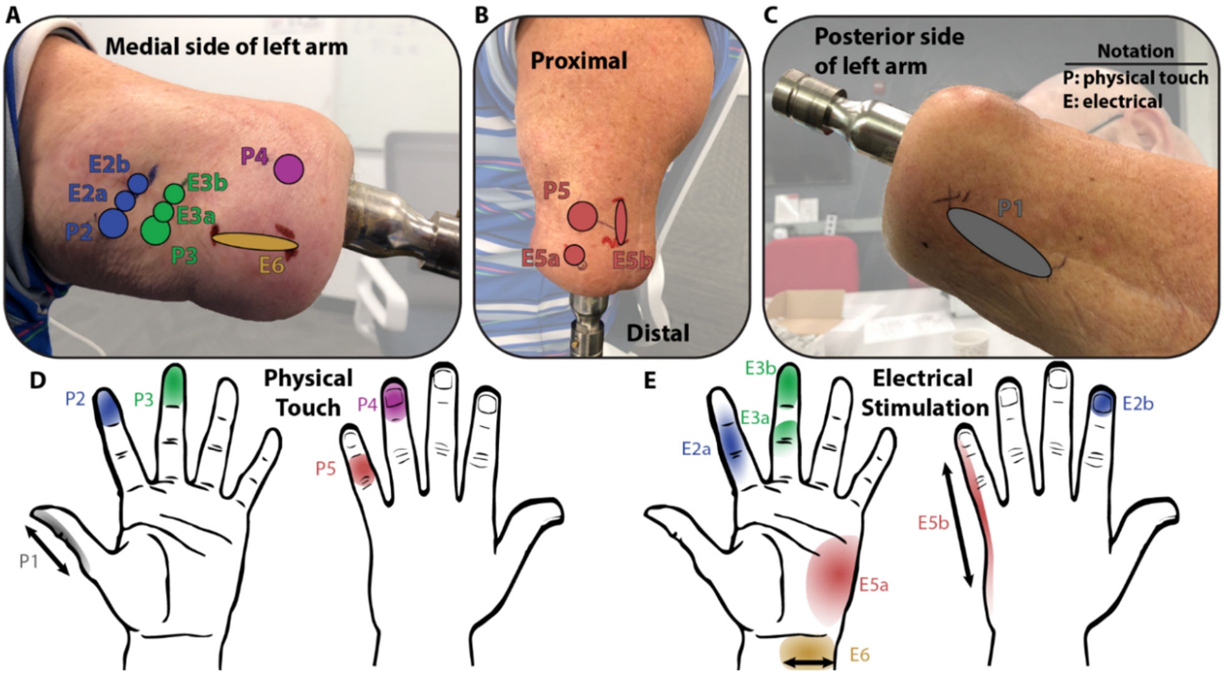

Figure 1: Phantom hand activation from physical and electrical stimulation. (A) Sensory stimulation sites on the medial portion of the arm that correspond to the index finger, middle finger, and wrist. (B) Little finger and (C) thumb sensory stimulation sites on the residual limb. Larger circles represent sites where physical touch activates the phantom hand, and the smaller circles represent sites that activate the phantom hand during tTENS. (D) Sensory activation in the phantom hand during physical touch of the corresponding sites on the residual limb. (E) Sensory activation in the phantom hand during tTENS.

regions, on the residual limb, of sensory activation in the phantom hand. Once a stimulation site on the residual limb was found, the participant used a marker to draw out the activated regions.

Targeted TENS was used to electrically activate underlying nerves in the residual limb to elicit sensory perceptions in the phantom hand. Sensory mapping was performed by scanning a 1 mm beryllium copper (BeCu) probe across the surface of the skin on the residual limb. The frequency (f) of electrical stimulation ranged from 2 – 4 Hz and the pulse width (pw) was 5 ms. The amplitude of the stimulation (I) ranged from 1.5 – 1.8 mA. We've validated the tTENS method in previous studies [7], [9] The locations that elicited sensory activation in the phantom hand were marked on the residual limb.

RESULTS

Sensory activation of the phantom hand is shown in Fig. 1. Locations on the residual limb that correspond to regions of the phantom hand are labeled in Fig. 1A-C. The sites on the residual limb that activate the phantom hand during physical touch are labeled with *P*, whereas the sites that respond to tTENS are labeled with *E*. The phantom hand activation for each stimulation site is shown in Fig. 1D-E. The participant reported that sensory stimulation was perceived like a pressure or a light touch and was localized to the phantom hand for both physical touch and tTENS.

The phantom thumb was only activated during physical touch (P1) whereas the palm and wrist were only activated during electrical stimulation (E5 and E6, respectively). The arrows next to P1 and E6 indicate that the participant could feel the physical touch (P1) or the tTENS probe (E6) moving within the sensitive region on the residual limb. The participant reported that these sensations were localized to the phantom hand.

The index and middle fingers were activated during both physical touch and tTENS. Further, the region of activation was similar for both modalities in the index finger, but differed slightly in the middle finger. For both index and middle fingers, the stimulation sites on the residual limb were different for the physical and electrical stimulations; however, they were relatively close to each other.

DISCUSSION & CONCLUSION

Based on our observations, the sites on the residual limb that are linked to activation of the phantom hand are different for physical touch and tTENS. That being said, we did observe that some of the locations, specifically for the index and middle fingers, are close in proximity. The fact that these stimulation locations are close could be indicative of the underlying sensory nerve fibers that respond to TENS being along the same nerve fascicle with fibers that innervate the skin at locations where physical touch elicits sensory activation in the phantom hand.

We believe that the physical touch sites on the residual limb are likely areas of the skin where sensory nerve fibers reinnervated superficially and thus produce action potentials as a result of physical manipulation. The underlying nerves in regions activated by tTENS are likely deeper in the soft tissue and are activated by the electrical pulses. It is reasonable to consider the possibility that the physical touch activation sites contain nerves reinnervated into the skin, and tTENS responsive sites are regions where nerve fibers or fascicles are close enough to the surface of the skin to allow electrical activation of the fiber or fascicle. The mechanical manipulation at reinnervated sites or where nerve fibers terminate likely causes the perceived sensation in the phantom hand. The tTENS sites on the residual limb are likely regions where electrical stimulation penetrates along the path of a fiber, eliciting the phantom sensory activation.

Because of the different mechanism of nerve activation (mechanical manipulation of reinnervated nerves and electrical stimulation of underlying nerve fibers or fascicles), it might explain why we didn't observe physical touch and tTENS sites being at exactly the same location on the residual limb. The force exerted on the skin by the TENS probe was likely not large enough to elicit mechanical activation of the reinnervated sites on the residual limb that corresponded to sensory activation of the phantom hand during touch.

Some regions, like the thumb and wrist are only activated by either physical touch or tTENS, respectively. The thumb responding to physical touch but not tTENS could be due to the underlying nerve fibers or fascicle innervating that location being too deep for the electrical stimulation to reach it. Similarly, the region of tTENS wrist activation could have an underlying nerve fascicle that is superficial enough to be activated by electrical stimulation, but the reinnervation occurs deeper in the soft tissue, thus preventing mechanical stimulation on the surface of the skin.

Every amputation case is different and each participant requires thorough sensory mapping to understand the perceived sensations in the phantom hand due to physical and electrical stimulation. Although we previously explored tTENS in multiple subjects [7], [9], every sensory map is different and varies between participants. As closed-loop prosthesis research continues to advance, it is important to explore and quantify the various forms of sensory stimulation modalities and resulting perceptions in amputees.

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A PORTABLE PROPORTIONAL CONTROL PROSTHESIS WITH HIGH-RESOLUTION DATA LOGGING

Mark Brinton[1], Elliott Barcikowski[2], Tyler Davis[3], Michael Paskett[1], Jacob George[1], Gregory Clark[1]

1. Biomedical Engineering, University of Utah, Salt Lake City, UT, USA; 2. Ripple Neuro LLC, Salt Lake City, UT, USA; 3. Neurosurgery, University of Utah, Salt Lake City, UT, USA.

ABSTRACT

This paper describes a portable, prosthetic control system that uses a modified Kalman filter to provide 6 degree-of-freedom, real-time, proportional control. We describe (a) how the system trains motor decoding algorithms and controls an advanced bionic arm, and (b) the system's ability to record an unprecedented and comprehensive dataset of EMG, hand positions and force sensor values. This technology enables at-home dexterous bionic arm use, and provides a high-temporal resolution record of daily use—essential information to determine clinical relevance and advance future research.

INTRODUCTION

Commercially available prostheses suffer from a variety of limitations, including: a limited number of pre-determined grips, temporal delay due to sequential inputs used to select grips, fixed output force (e.g., from classifier algorithms), extensive training that takes days to weeks, and non-intuitive methods of control (e.g., inertial measurement units on arm or feet) [1]–[3]. Advanced control of multiple degrees-of-freedom, and the training associated with them, are not always amenable to deployment on portable systems with limited computational power. However, a Kalman filter [4], modified with non-linear, ad-hoc adjustments [5], [6], can provide a computationally efficient approach to proportionally and independently control multi-degree-of-freedom prostheses.

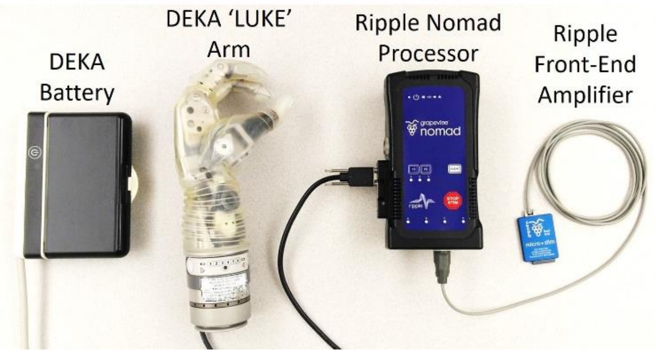

Figure 1 – Portable take-home system includes the DEKA LUKE Arm and battery, the Ripple Nomad neural interface processor and battery (hidden) and a front-end amplifier (amplifier for surface EMG shown here).

In a previous publication we demonstrated supervised at-home use of a portable, prosthetic control system that relied on a modified Kalman filter to provide 6 degree-of-freedom, real-time, proportional control [6]. Here, we describe this system including: (a) how it can be used to train motor decoding algorithms and control an advanced bionic arm; and (b) its ability to record an unprecedented dataset of electromyography (EMG), hand positions and force sensor values. This technology constitutes an important step toward the commercialization of dexterous bionic arms by demonstrating at-home use of proportional control, multi-degree-of-freedom prostheses and recording high-temporal resolution data describing the arm use.

METHODS

Design Considerations

A portable take-home system designed to research advanced bionic arms should meet several criteria for optimal performance and data collection: (a) the system must accurately and efficiently control the prosthetic arm; (b) training of the prosthetic arm must not be too long or burdensome to prevent its daily use; (c) high temporal resolution data should be stored automatically so that researchers can study at-home use without influencing the users with in-person observation; and (d) the system must be easy to use and allow the user to adjust control preferences.

Hardware and Signal Acquisition

The components of the portable system are shown in Figure 1, including: (a) the 6 degree-of-freedom DEKA LUKE Arm (Manchester, NH) and its 13 force sensors (0 to 25.5 N) and rechargeable battery; (b) the Nomad Neural Interface Processor (Ripple Neuro, Salt Lake City, UT) with a more than 4-hour, rechargeable battery, a 500 GB disk storage and up to 512 channels of data acquisition; and (c) the front-end amplifier (Ripple Neuro, Salt Lake City, UT) which filters (15 to 375 Hz bandpass; 60/120/180 Hz notch) and amplifies 1-kHz sampled EMG data. Surface EMG in intact participants was recorded with a Micro + Stim front-end, and implanted EMG in the amputee participant was recorded with an active gator front end. Ripple also provided firmware with the Nomad for data acquisition (EMG at 1 kHz; LUKE Arm positions and force sensors at 33 Hz), communication with the LUKE Arm (CAN bus protocol), ability to start and stop compiled decoding algorithms via external buttons, and WiFi communication to interface with external devices. The Nomad runs Linux 8 (jessie) environment with an Intel® Celeron™ processor (CPU N2930) at 1.83 GHz with 2 GB RAM. Algorithms were converted to C using MATLAB® Coder and compiled for stand-alone use on the portable Nomad.

Table 1: Computational times required for training and testing (running) the steady-state, modified Kalman filter.

Process	Computation Time
Training:	
Data collection	252 s
Channel Selection	198 s
Train Steady State Kalman Filter	0.7 s
Total Time	7.5 min
Testing:	
Update Positions	< 1 ms

Training, Feature Calculation and Motor Decoding

The Kalman filter presented by Wu et. al [4] was modified with external, ad-hoc thresholds as described in George et. al [6]. To improve stability and reduce the effort required to sustain grasping movements, a latching filter was also applied to the output of the modified Kalman filter [5]. Training the modified Kalman filter first requires the user to mimic a computer-controlled prosthetic arm as it cycles through several movement trials for each degree-of-freedom. Features were then calculated for each differential EMG pair (496 total from 32 single-ended electrodes) by taking the mean-absolute value of a moving 300-ms window. Using the movement kinematics and the EMG features, the compiled algorithm recursively chose the 48 most-descriptive features using the Gram-Schmidt forward selection algorithm [7] and then trained the Kalman filter matrices [4].

Human Subjects

Eight EMG leads (Ripple Neuro LLC; Salt Lake City, Utah, USA) with 4 contacts each, and a ninth lead with an electrical reference and ground, were implanted in the lower-arm extensor and flexor muscles of a trans-radial amputee as described elsewhere [6]. Intact individuals used the portable system with a bypass socket [8] and a custom-made neoprene sleeve with 32 surface EMG electrodes, plus 1 reference and 1 ground (George et al., MEC, 2020). All experiments and procedures were performed with approval from the University of Utah Institutional Review Board.

RESULTS

Three external buttons were employed to create a simple user-friendly interface. Pressing the first button initiated a new training session. The second button initiated a previously trained and compiled motor decoding algorithm so that the user could have on-demand control of the arm. A third button was used to toggle between position or velocity control modes or to freeze a degree-of-freedom at a desired position.

Figure 2 - Two-handed activities-of-daily living in the home using a bypass socket and the portable system: (a) using scissors, (b) donning a sock and (c) folding a towel.

The system was trained in about 7.5 minutes—including a movement mimicry session (252 sec) and the subsequent selection of the optimal channels and computation of the steady-state Kalman filter matrices (about 199 sec) (Table 1). Training data included 4 trials for each of the thumb, index, middle/ring/little and wrist flexion and extension; thumb adduction and abduction; wrist pronation and supination; and grasping and extending all digits together. The trained Kalman filter was automatically saved to a log file and could be recompiled onto the Nomad as a stand-alone application for on-demand use (e.g., the second external button). This was accomplished over the Nomad's wireless network using a laptop and required less than 30 seconds.

Prior to use, the steady-state Kalman gain matrix (K) was calculated by iteratively running the filter until the fluctuations in every value of the gain matrix were less than $1x10^{-6}$, reaching steady state after about 25 ms. With the gain (K), the observation (H) and the state-transition (A) matrices, a steady state matrix (Γ) was then calculated:

$$\Gamma = A - K * H * A \tag{1}$$

Thus, new position predictions ($\hat{x}$) were calculated with only two matrix multiplications involving the previous positions and the 48 selected EMG features (z):

$$\hat{x}_{new} = \Gamma * \hat{x}_{previous} + K * z \tag{2}$$

This simplification avoided a computationally expensive matrix inversion required by the recursive algorithm. Consequently, the time required to predict new positions and update the prosthesis was, on average, less than 1 ms, far below the update loop speed of 33 ms (see Table 1). The portable system was used at home to perform two-handed tasks with both intact participants (Fig. 2) and a trans-radial amputee [6].

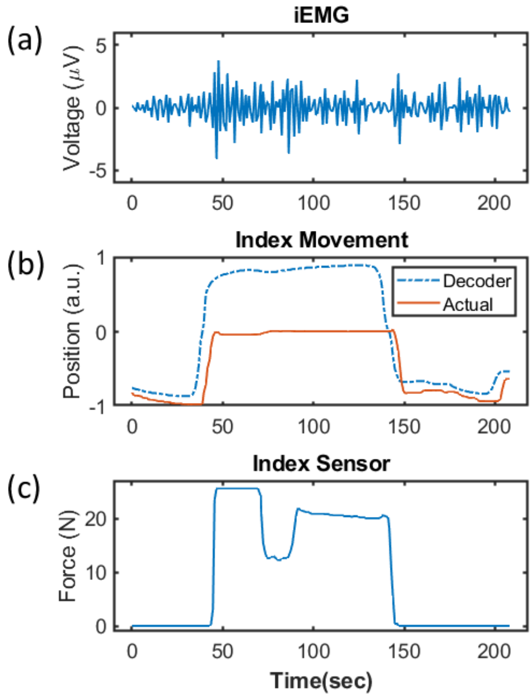

Figure 3 - (a) A differential implanted EMG channel (iEMG, at 1 kHz), (b) motor decoding algorithm estimates and actual arm positions (at 30 Hz) and (c) force sensor values (at 30 Hz) recorded while a trans-radial amputee grasped, held and released an orange.

Comprehensive EMG (sampled at 1 kHz), arm positions and arm forces (both sampled at 30 Hz) were stored on the Nomad while a trans-radial amputee grasped, held and released an orange (Figure 3). Figure 3 shows one differential pair of the implanted EMG (iEMG) and the index finger positions and force. A grasp occurred when the index position moved from -1 to +1. During the grasp, the difference in position between the motor decoding algorithm's estimated and the actual positions occur because the object prevents the finger from flexing to its full extent (Fig. 3b). This caused a dramatic increase in force (Fig. 3c). Data from the 32 EMG channels, 6 arm positions and 13 force sensors are saved at a rate of 250 MB/hour in an '.hd5' format. As a result, the 500 GB capacity of the Nomad can record nearly 2000 hours of arm use.

DISCUSSION

We have shown that a modified Kalman filter can be trained in about 7.5 minutes to proportionally control 6 independent degrees-of-freedom using the Nomad portable processor. The portable system has been used in the lab and at home by intact persons, as well as by a trans-radial amputee to perform tasks not possible with his commercial prosthesis [6]. Even with an ordinary microprocessor, position updates were generated much faster than the 33-ms loop speed, providing the users with real-time control. The portable system also stores EMG, position and force data with unprecedented temporal resolution. This comprehensive dataset will be crucial for fully understanding how proportional control algorithms are used during unsupervised at-home use.

Figure 3 highlights how the comprehensive data recorded by the Nomad reveals complex interactions between the various degrees of freedom for improved control. The stable index finger position implies that the amputee held the orange with a fixed grasp from pick up to release; however, the force data revealed a dip in force during this same period. Close inspection of the position data of the opposing thumb (not shown) also shows that a subtle readjustment occurred to improve the grasp stability. Because degrees of freedom are coupled together during object manipulation, the connection between each degree of freedom must be considered. Due to complex regional pain syndrome, the trans-radial amputee in this study had kinesiophobia and had not used his hand for several years prior to amputation. As a result, the recorded EMG signals were often very weak (Figure 3a). However, with these weak signals, the portable system and motor decoding algorithms still provided the participant the functional control necessary to complete common daily tasks in his own home.

Rich datasets like this will help researchers study at-home, unsupervised prosthesis use; improve motor decoding algorithms and training paradigms [9] by understanding the types of grasps and degrees of freedom commonly used; understand when mastery of prosthetic control occurs and when interventions might be applied or lifted; better describe noise encountered in real-world environments and design features and algorithms that reduce its influence on motor performance; and address many other unanswered questions about at-home use of advanced upper-limb prostheses. These rich datasets will also enable future at-home trials to study the benefits and use of high-resolution sensory feedback from intraneural electrical stimulation—a feature soon to be added to the portable system.

The most computationally demanding aspect of training was performing Gram-Schmidt forward selection to choose the 48 most useful features out of the 496 differential pairs. Despite taking considerable time up front, this down-selection method has several advantages [7]. First, choosing the features up-front enables fast loop speeds (below 33 ms) by eliminating the need to calculate complex features (e.g., principal components) or even all 496 differential EMG features during each update cycle. Second, forward selection recursively chooses independent and informative features using orthogonality reduction and correlation with the training kinematics. This ensures that each selected feature describes kinematics and not uncorrelated noise. Refined movements, the hallmark of proportional control algorithms, account for little variance and could be inadvertently discarded using techniques agnostic to the training kinematics. Finally, orthogonalization in the forward selection algorithm avoids redundant features and singularities.

A key feature of the portable system is that the time from powering the system to having real-time proportional control is less than 8 minutes. The amount of time required to both collect training data by mimicking arm movements and train the motor decoding algorithm are related to the number of trials for each mimicked movement. In this work, and published elsewhere, an amputee familiar with the training process only trained with 4 trials on each degree-of-freedom and a grasp and extension of all digits. With this training, he was able to control the arm in the lab and perform tasks not possible with his commercial prosthesis at home [6]. A less experienced user may require training with more trials; however, even if a naïve user requires twice as many trials the total training time (mimicry and computation) is still under 15 minutes.

In its current form, the portable system is only able to communicate over a CAN bus python socket with the DEKA LUKE Arm. However, other custom communication sockets could be designed to communicate through the micro D-sub, USB or Bluetooth connections available to Nomad for proportional control of and data logging from other prosthetic limbs.

In the future, the portable system will also include sensory stimulation for haptic feedback in response to the forces experienced by the prosthesis. Ultimately, this system will be used in upcoming take-home clinical trials to record high-resolution data and study advanced, proportional control algorithms and sensorized prostheses in trans-radial amputees.

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TOUCH FEEDBACK AND CONTACT REFLEXES USING THE PSYONIC ABILITY HAND

Aadeel Akhtar, Jesse Cornman, James Austin, Dhipak Bala
PSYONIC, Inc.

ABSTRACT

The PSYONIC Ability Hand is a commercially available multiarticulated prosthetic hand with six degrees of freedom and sensorized digits. Through using contact reflexes and vibration feedback, users can grasp delicate objects without damaging them. We show results that two subjects successfully grasp hollow eggshells and fragile cups statistically significantly more often when provided with contact reflexes and touch feedback.

INTRODUCTION

The Ability Hand

PSYONIC has developed the commercially available Ability Hand—a compliant, robust, sensorized prosthetic hand to be used by people with upper limb amputations. The Ability Hand is:

- Multiarticulated – all five digits flex/extend and the thumb rotates both electrically and manually
- Robust – compliant fingers allow the hand to withstand blunt force impacts to the fingers
- Lightweight – 460 g, carbon fiber palms make the hand light and strong
- Fast – using brushless motors with field-oriented control, the fingers can close 90 degrees in 200 ms
- Waterproof – IP64 waterproof rating, enabling washing the hand in water
- Sensorized – pressure from the fingertips, fingerpads, and lateral edges maps to a vibration motor

The Ability Hand uses a standard electronic quick disconnect and integrates with commercially available control systems (e.g. Coapt Pattern Recognition, OttoBock/RSL Steeper myoelectrodes, etc.). Apple and Android phone apps are available to configure the hand over Bluetooth as well as make firmware updates. USB-C charging allows the hand to be fully charged within one hour.

Fig. 1 The Ability Hand attached to a socket

Sensory Feedback

Poor manipulability due to the lack of sensory feedback is a leading cause of prosthesis abandonment [1-2]. While body-powered prostheses can give users some sensory feedback, these devices are limited in achievable grasps and can cause overuse injury in the shoulders of the user. There are several functional advantages to providing sensory feedback in a multiarticulated prosthesis, including contact detection and body self-identification [3]. An external study by Matulevich et al. [4] shows that users could grasp foam, crackers, and hollowed eggs statistically significantly faster (between 1.4x-3.3x) when using contact detection from pressure sensors on a prosthetic hand. Another external study by Berke et al. [5] showed users performing tasks more than 15 seconds faster on average when provided with contact detection.

In the Ability Hand, all five digits can be sensorized with four pressure sensors in each digit. The index and little fingers have pressure sensors on the distal fingertip, the fingerpad, and two on the outer lateral edges. The thumb, middle, and ring fingers typically have pressure sensors on the distal fingertip, the fingerpads, and one on each lateral side of the digit. These pressure sensor locations were chosen due to their increased likelihood of contacting objects. The sensor providing the highest pressure value is mapped to a vibration motor whose amplitude changes with the pressure applied.

To test the efficacy of the sensory feedback, we recruited two volunteer subjects. The first subject, S1, was a male, age 42, with a right proximal below-elbow amputation. The second subject, S2, was a male, age 78, with a left distal below-elbow amputation. S1 was fitted with a commercial muscle pattern recognition system developed by Coapt that we integrated to use with PSYONIC's hand. S2 used a custom linear transducer mechanism developed by PSYONIC that uses shoulder movements to control opening and closing the hand.

Subjects S1 and S2 were asked to use the hand at home for 1 week. Immediately prior to and after the home trial, both subjects participated in two experiments: 1) a cup grasping task, and 2) an eggshell cracking test. All methods were approved by IRB #13920 at the University of Illinois at Urbana-Champaign. Subjects also consented to images and videos to be taken during the experiments. Preliminary experiments were performed in Akhtar et al. [6].

In the cup grasping task the subjects were asked to grasp ten empty plastic cups. The distance between the outer tips of the index finger and thumb was measured to determine the amount of deformation of the cup. This process was repeated over 4 conditions: 1) with Touch Feedback and with Visual Feedback, 2) without Touch Feedback and with Visual Feedback, 3) with Touch Feedback and without Visual Feedback, and 4) without Touch Feedback and without Visual Feedback. The order of the conditions was randomized. These conditions were selected to observe differences in grasping performance when providing touch feedback, both with and without visual feedback.

For the eggshell cracking test participants were asked to grasp ten hollowed eggshells without cracking them. We recorded the number of eggshells cracked. Again, the process was repeated under the same four conditions as the cup grasping task. When providing touch feedback to subjects, a contact reflex was implemented in the hand that caused the hand to automatically stop when contact with the object was made. Fig. 2 shows a typical pressure sensor reading when grasping a hollowed eggshell.

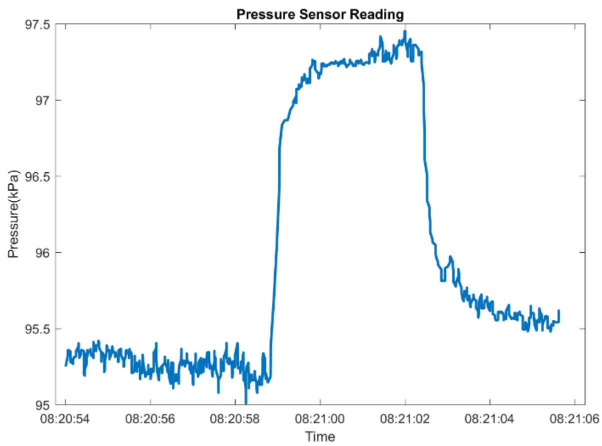

Fig. 2 Reading from pressure sensor on the finger pad of the distal index finger when Subject S1 grasped an eggshell.

Results from Subjects S1 and S2 across both sessions are given in Table I for the cup grasping task and Table II for the eggshell cracking test. For the cup grasping task, there was a statistically significant difference between feedback conditions as determined by a two-way repeated measures ANOVA ($F(3,3) = 567.7$, $p < 0.0005$). Post-hoc tests revealed that the touch feedback conditions (with or without visual feedback) statistically significantly outperformed both conditions without touch feedback ($p<0.05$). Consequently, we conclude that by providing touch feedback with contact reflexes users deform the plastic cup significantly less. There were no statistically significant differences between sessions, and the session had no significant effect on the condition.

Table I Results from cup grasping task

	Session	Touch, Visual (mm)	Touch, No Visual (mm)	No Touch, Visual (mm)	No Touch, No Visual (mm)
S1	1	80.3	79.5	37.4	39.4
	2	78.7	82.2	51.4	48.9
S2	1	77.3	80.2	38.6	38.9
	2	87.3	89.5	49.5	54.3
Grand Mean		**80.9**	**82.9**	**44.2**	**45.4**

For the eggshell cracking test, there was a statistically significant difference between feedback conditions as determined by a two-way repeated measures ANOVA ($F(3,3) = 21.63$, $p = .016$). There was no statistically significant differences between sessions, and the session had no significant effect on the condition. Again, touch feedback with contact reflexes resulted in better performance, with less eggshells cracked compared to when no touch feedback with contact reflexes was given (with or without visual feedback).

Table II Results from eggshell cracking test

	Session	Touch, Visual (# cracked)	Touch, No Visual (# cracked)	No Touch, Visual (# cracked)	No Touch, No Visual (# cracked)
S1	1	0	2	7	9
	2	0	3	6	6
S2	1	0	0	7	7
	2	2	1	8	6
Grand Mean		**0.5**	**1.5**	**7**	**7**

Fig. 3 shows images of the Subject S1 performing the eggshell cracking test. When touch feedback with contact reflexes was turned on, the subject could easily grasp the eggshell without cracking it, even while blindfolded. When touch feedback with contact reflexes was turned off, the subject usually cracked the eggshell, even when he could see it. Fig. 4 shows Subject S2 successfully grasping the eggshell while blindfolded when receiving touch feedback with contact reflexes.

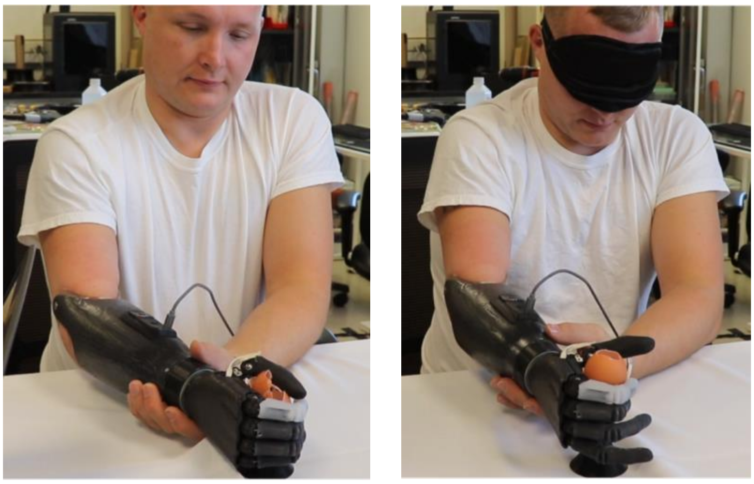

Fig. 3 Subject S1 cracking an eggshell when not receiving touch feedback while seeing the eggshell (left), but successfully grasping the eggshell when receiving touch feedback while blindfolded (right).

Fig. 4 Subject S2 successfully grasping the eggshell when receiving touch feedback while blindfolded.

Qualitative feedback from the subjects after the home trials was positive. Subject S1 reported he mostly wore the hand during work. Common tasks included holding drinks, driving, shaking hands, and sweeping. He liked the light weight of the prosthesis as well as the bionic look. Subject S2 liked the fact that our hand could work with both off-the-shelf myoelectric systems and a linear transducer system. He used a linear transducer system for the trial that he found to perform vastly better than a myoelectric system. He used this mainly to grasp glasses to drink from, for exercising on a stairmaster, and for assistance in typing (e.g. holding down the shift button on a keyboard). For improvements, he expressed that multiple settings for the pressure sensor contact reflexes would be helpful, as some objects require tight grips while others require delicate grips.

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TOWARDS OBJECTIVE ASSESSMENT OF OWNERSHIP OVER A PROSTHESIS

Daniel H. Blustein[1,2], Noah B. Mesa[2], Erin S. Kuylenstierna[2], Kelley E. Parsons[2], Cierra J. Stiegelmar[3], Jacqueline S. Hebert[4], Ahmed W. Shehata[4]

[1]Department of Psychology, Rhodes College. [2]Neuroscience Program, Rhodes College, Memphis, TN, USA. [3]Faculty of Science, University of Alberta. [4]Division of Physical Medicine and Rehabilitation, University of Alberta, Edmonton, AB, Canada.

ABSTRACT

Conventional motor assessments provide limited actionable information to prosthetic clinicians and engineers. Recent work has sought to develop objective ways to measure psychological aspects of a person controlling a prosthesis to develop more powerful motor assessment tools. One area of emphasis has been to develop a way to objectively measure device ownership, a key component of embodiment. Assessment of ownership has historically been limited to subjective questionnaires but here we use a spatial interference reaction time task, the crossmodal congruency task (CCT), to objectively assess this key factor in supporting prosthesis use. We improve the CCT protocol to increase its usability. We aim to establish a causal link between 'device ownership' and the crossmodal congruency effect, a correlational link observed in previous work. In this paper we summarize our efforts to develop a comprehensive platform to assess ownership and share results from an initial study.

INTRODUCTION

Emerging prosthetic devices using peripheral nerve interfaces [1], [2], targeted reinnervation [3], [4] and non-invasive control and feedback strategies [5], [6] show promise. However, the methods used to assess these technologies often provide limited information. Performance measures, such as the Box and Blocks Test [7], the Southampton Hand Assessment Procedure [8] and the Jebsen Hand Function test [9], provide little mechanistic insight with results that can be distorted by interacting compensatory movements [10]. Functional assessments like the Assessment of Capacity for Myoelectric Control [11], although shown to be reliable [12], rely on subjective scoring provided by trained raters. Recent efforts in the development of motor system assessments have focused on objective measures that are theoretically-grounded in neuroscientific and psychological principles [13]. Quantifying aspects of a prosthetic system that are involved in control of an intact limb may aid in identifying deficiencies in the engineered systems that might *explain* differences in observed motor performance. The goal is to use assessments to inform, target and customize device improvements to try to better mimic their biological system counterparts.

Recent efforts to develop objective assessments have focused on measuring the psychological factors that are involved in a motor system. One goal in engineering a prosthetic device is to convey to the user a sense of embodiment [14]. That is, the device is felt as an integrated part of one's body [15]. Although there remains some debate about what psychological aspects contribute to the sense of embodiment, it is thought that both a sense of ownership and a sense of agency (or control) over the device are required [16], [17]. Much work has been done to assess ownership, agency, and embodiment overall but these studies typically rely on subjective questionnaires [4], [18]. Furthermore, these studies are often correlational, lacking direct experimental manipulations to identify causal links between factors contributing to embodiment. When experimental manipulations are undertaken, they usually focus on one aspect of embodiment (e.g. ownership [19] or agency [20]), and not their interaction.

We have undertaken a series of studies [21] to explore embodiment using a standardized simulated prosthesis system. We aim to simultaneously assess the sense of agency and the sense of ownership using objective measures. In this study we focus on ownership assessment. But why is it important to measure ownership and agency with respect to prosthetic devices? We argue that if a device is more incorporated into one's body image by feeling (ownership) and moving (agency) like one's own biological limb, it will be more functional and more useful. We anticipate that increases in ownership and agency will lead to better motor performance, reduced user frustration, increased device use and reduced rates of device rejection which has been shown to be a key roadblock in prosthetic device implementation [22].

Our recent work has developed objective ways to measure the sense of ownership using an adapted crossmodal congruency task (CCT) [23], [24]. An increase in ownership is correlated with an increase in the reaction time (RT) difference between congruent (aligned) and incongruent (misaligned) sensory stimulation. This RT difference is called the crossmodal congruency effect (CCE) score. Here we present an improved protocol for assessing ownership using a simulated prosthesis. First, we highlight the improvements we have made on our previous work and that of Marini et al. [25]. Then we describe results from an initial study and ongoing experiments to validate this new experimental platform. Finally, we present our upcoming experimental plans to better understand the dynamics of device ownership and its interaction with the sense of agency.

IMPROVING THE CROSSMODAL CONGRUENCY TASK

We first sought to develop an improved objective ownership assessment. Across individuals, we observe high variability in CCE scores [26]. This would imply that within-subject assessment may be more informative (as in [25]). However, due to a practice effect observed with the repeat use of the CCT [26], we expect changes in CCE scores with repeat task completion. Therefore, we looked to improve the implementation of the CCT by reducing inter-individual variability to allow for the use of between-subjects designs in future experiments.

One potential reason for the high variability in CCE scores was due to the random trial order presented to participants. In our previous work [23], the stimulus condition (congruent vs. incongruent, and location) was randomized independently for each trial. Typically, half of testing trials are congruent and the other half incongruent. However, if during initial practice and testing trials the actual percentage congruent were not 50% (which is likely in small samples of random trials) we might see different learning dynamics occur. Congruency expectation might lead to different responses and variable CCE scores, early model learning that could persist throughout testing. When presented with different percentages of congruent trials in other psychophysics tasks, we see significant changes in the RT differences between congruent and incongruent stimuli [27].

We analyzed previously collected CCT data [26] to determine if a congruency sequence effect is present in CCT results. RT data on correctly discriminated trials were sorted into four groups representing the possible congruency combinations for each pair of trials: Congruent then congruent (CC); Incongruent then congruent (IC); Incongruent then Incongruent (II); and Congruent then Incongruent (CI). We calculated the RT z-scores for each subject and ran a one-way ANOVA with Bonferroni-corrected post-hoc comparisons for the results in each trial congruency pairing.

Trial pairs for which congruency is switched show slower RTs than when congruency is consistent between two trials (Figure 1). Although this observation was not statistically significant as determined by a one-way ANOVA, the trend in Figure 1 shows a congruency sequence effect. Therefore, we updated the CCT protocol to use pre-generated pseudo-random sequences of trial conditions as is common practice with studies using different interference tasks like the Stroop Test and the Flanker Task [27]. We generated pseudo-random sequences of test stimuli that ensured each paired order of trials appeared equally often. This ensured no more than 4 trial types (e.g., 4 congruent trials) could occur in a row. We generated 4 different test sequences of 64 trials each, and the order of these 4 sequences is randomized for each participant. We similarly generated 3 practice sequences of 8 trials each and the order of their presentation was randomized during the practice phase.

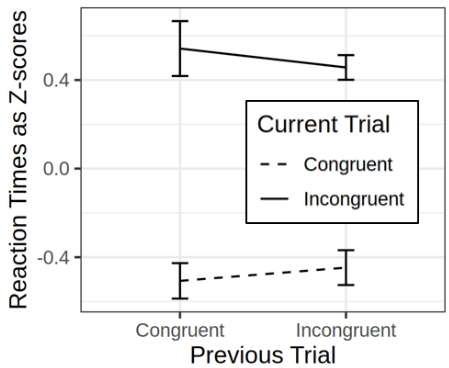

Figure 1. RTs for different congruency sequence pairs during the CCT.

IMPROVING THE PROSTHESIS SIMULATOR SETUP

One study using a simulated prosthesis with CCE assessment used a fixed prosthesis mounted to a table with the able-bodied user controlling hand open and close [25]. This approach could potentially limit the degree of embodiment attainable and provides for less realistic movements than the freely moving simulated prosthesis we use here. Our previous work in this area used a heavier simulated prosthesis [23], [28]. By reducing the mass, from 1.43kg to 0.66kg, we expect reductions in EMG signal noise and user fatigue. We also use an improved mechanotactile tactor [21] to apply force feedback on the user's fingertips, driven proportionally by signals from force sensitive resistors embedded in the index finger and thumb of the prosthesis. We ensured that with no force applied to the finger/thumb sensors, there was no contact between the tactor and the user's skin. This approach was not taken in [23] and may explain some unexpected results in that study.

TEST PLATFORM VALIDATION

We ran an initial study to determine if the newly developed system would operate consistently and lead to consistent ownership assessments. Written informed consent per Rhodes College IRB oversight was obtained for each participant. After a participant donned the simulated prosthesis and MyoBand (see [21] for hardware details), EMG settings were calibrated. Two of the eight electrodes were used: the one with the highest signal-to-noise ratio (SNR) during wrist flexion and the one with the highest SNR during wrist extension. The participant's baseline EMG activity and maximum voluntary contraction (MVC) activity were measured and used to set the activation threshold and gain, respectively, for both electrodes. The threshold was set at about two times the baseline EMG activity level, and the gain was adjusted to map the prosthetic hand velocity from the threshold (V_0) to the MVC level (V_{max}). Wrist flexion was mapped to hand close, and wrist extension to hand open.

Participants trained by grasping an object with the prosthesis (right hand) and moving it to the left over a barrier 14.5 cm high. The object was dropped after crossing barrier, the experimenter retrieved the object and placed it back on the right side of the barrier for the user to start the next movement. Two participants completed 30 training movements with a break halfway through for two different training conditions. In the voluntary control condition, the person's EMG activity controlled the hand. In the involuntary control condition, the experimenter controlled the opening and closing of the hand with the participant matching the hand movement with their EMG activation. After training, participants completed a questionnaire assessing embodiment, ownership, agency and localization. Table 1 shows the ownership statements to which users indicated their agreement on a continuous scale [from Strongly Disagree to Strongly Agree] that was converted into a -5 to +5 score.

Switching between involuntary and voluntary control would affect agency as shown in a concurrent study [29], but not ownership. We observed similar ownership scores across both conditions in both participants (Figure 2).

Table I. Ownership post-training questions [31]

OWNERSHIP
I felt like I was looking directly at my own hand, rather than at a prosthetic hand.
I felt like the prosthetic hand was my hand.
I felt like the prosthetic hand was part of my body.
I felt like the prosthetic hand belonged to me.
It seemed like the prosthetic hand became to resemble my real hand.
OWNERSHIP CONTROL
I felt like my real hand was turning rigid.
It seemed like I had more than one right hand.
The prosthetic hand started to change shape, colour, and appearance so that it started to (visually) resemble my hand.

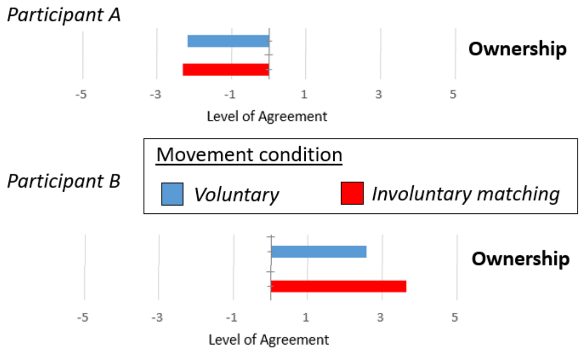

Figure 2. Average response on ownership questions for each participant and condition.

DISCUSSION

This study's results suggest that the updated simulated prosthesis is a suitable platform to test questions related to device ownership. The device is lighter than those used previously and we observed consistent ownership results via questionnaire assessment in an initial study. Ongoing studies will attempt to validate the objective psychophysics-based CCT by investigating the correlation between questionnaire results and CCE scores. We are also seeking to determine the training duration necessary to elicit device ownership. To further validate the CCT as an ownership assessment we can experimentally manipulate the level of ownership (e.g. by adding feedback delay), and then observe the sensitivity of the CCT response compared to the sensitivity of questionnaire results.

We provide some evidence that the CCE is subject to the congruency sequence effect like the Stroop task [30]. For future statistical analysis, we will use a multi-level mixed effects design to further control cofounding variables and quantify order effects more precisely. Additionally, we will run a CCE experiment which varies the percentage of congruent and incongruent trials to see if the congruency sequence effect can be mediated by participant expectations of future trials.

This is ongoing work intended to characterize embodiment development during prosthesis use. Our next step is to characterize the training duration necessary to elicit ownership with our prosthesis simulator system. Previous studies using simulated prostheses have shown quite varied durations of training necessary from about an hour [23] to 30 hours [25]. We expect to observe embodiment with much shorter durations of training because we are using a dynamic prosthesis simulator, unlike [25], that is lightweight (unlike [23]) and we have adopted a new protocol aimed to reduce the inter-individual variability in CCE score results.

In our study establishing the relationship between training duration and ownership for this device, we will also implement the CCT along with the questionnaires. Both previous studies using CCE assessment with robotic hands [23], [25] did not correlate CCE results with results from established questionnaires. We expect to see ownership increase with increased training duration and expect a positive correlation between questionnaire results and CCE scores.

Once the validation studies are complete, we can test various questions related to prosthesis ownership and how this concept interacts with other psychological aspects of device use. For example, we can look at how emerging feedback systems affect device ownership. We will also investigate the relationship between ownership and agency in prosthesis use. Previous work has focused on one of these aspects alone, or relied on subjective questionnaires. Along with concurrent work developing a robust measure of agency [29], we can objectively assess both ownership and agency at the same time in the same platform.

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TOWARDS QUANTIFYING THE SENSE OF AGENCY AND ITS CONTRIBUTION TO EMBODIMENT OF MYOELECTRIC PROSTHESES

Cierra J. Stiegelmar [1], Daniel H. Blustein [2], Jonathon W. Sensinger [3], Jacqueline S. Hebert [4], and Ahmed W. Shehata [4]

[1]Faculty of Science, University of Alberta, Edmonton, AB, Canada. [2]Department of Psychology, Rhodes College, Memphis, TN, USA. [3]Institute of Biomedical Engineering, University of New Brunswick, Fredericton, NB, Canada. [4]Division of Physical Medicine and Rehabilitation, University of Alberta, Edmonton, AB, Canada

ABSTRACT

Myoelectric technology has the potential to improve prosthetic device functionality. However, device rejection rates remain high, an observation partly attributed to a lack of sensory feedback and difficult control strategies in these devices. Sense of agency, or feeling of control over one's actions, may be able to address these high rejection rates, but existing studies tend to rely on subjective questionnaires to study this experience. Evidence suggests that intentional binding, the compression of the perceived time interval between a voluntary action and its sensory effect when an individual feels in control, may be a quantifiable correlate of the sense of agency. However, existing intentional binding protocols are susceptible to expectation bias and are attentionally demanding for participants. Psychometric assessment tools, such as two-alternative forced choice, may be able to quantify this subjective experience while avoiding bias and attentional demand. In this work, we developed an experimental protocol that uses a psychometric assessment method, namely a two-alternative forced choice paradigm, to study intentional binding and the sense of agency. Here we present preliminary results from 2 able-bodied participants using a myoelectric simulated prosthesis fitted with mechanotactile feedback during voluntary and involuntary control conditions for a grasp-and-release task. These results show that responses to sense of agency questionnaire items are affected by voluntary and involuntary control of a prosthesis.

INTRODUCTION

Researchers report high rejection rates among advanced myoelectric prosthesis users [1] with mean adult and pediatric rejection rates of 23% and 32%, respectively [2]. Lack of sensory feedback [3], [4] and difficult control strategies [5], [6] are both argued to play a part in myoelectric prosthesis rejection. These factors may reduce an individual's sense of embodiment over a device, which may also contribute to device rejection [7].

Embodiment, which refers to the feeling that occurs when one experiences their body as their own and that they exist within it [8], [9], is made up of three interrelated factors: localization, ownership, and agency. Localization refers to one's assumption of where their body exists in space [10], [11]. The sense of ownership describes the feeling when one experiences their body as belonging to oneself [11]. Agency occurs when agreement between sensory predictions and sensory experiences leads to a feeling of control over one's actions and the resulting impact on the surrounding environment [12]. The experience is stronger when predicted sensory consequences of a voluntary motor action match the actual sensory consequences of the action [4]. Agency likely has great implications for prosthesis use because it depends on sensory information and certainty of control, which are factors that contribute to prosthesis rejection [3], [6]. In fact, increasing the sense of control may improve prosthesis acceptance [4].

This sense of control (agency) is commonly investigated using questionnaires, which provide valuable insight into the experience [13], [14]. However, the subjectivity in questionnaire response systems can introduce bias and limit comparison of results. A quantitative and objective investigation of psychophysical phenomena would allow for unbiased data analysis and comparison. When used alongside subjective questionnaires, it may lead to a more holistic understanding and informed approach to prosthesis technological development and training methods.

Intentional binding (IB) refers to the compression of the perceived temporal interval between action and effect when an individual feels in control of their actions, which may serve as a quantifiable correlate of the sense of agency [15]. Temporal estimation procedures can be implemented to quantify and compare a participant's estimated action-effect intervals for voluntary and involuntary control. In these experiments, involuntary control is used as a baseline value in which participants have no sense of agency over the movement. However, reporting methods used in existing studies are susceptible to response bias or are attentionally-demanding [16]. The use of psychometric approaches to quantify IB may mediate these issues.

One psychometric assessment method, known as a two-alternative forced-choice (2AFC) task, presents participants with two stimuli that differ by a specific parameter. The participant is asked to consistently identify and select the target stimulus out of the two stimuli. A correct response indicates that they are able to perceive the difference between the two stimuli. This procedure uses an adaptive approach to determine the level of difference between the two stimuli presentations at which the participant is able to indicate the correct target stimulus with only 50% (chance) accuracy [6], [17]. This method can be applied to IB research by quantifying a participant's perceived action-effect intervals for voluntary control with respect to involuntary control.

Here we developed an experimental protocol that uses a psychometric assessment method to objectively quantify the sense of agency, by determining a participant's perceived action-effect intervals for voluntary control with respect to perceived action-effect intervals for involuntary control. In order to ascertain the validity of this protocol, we first had to determine the influence of voluntary and involuntary control on sense of agency with a commonly used questionnaire [22].

MATERIALS AND METHODS

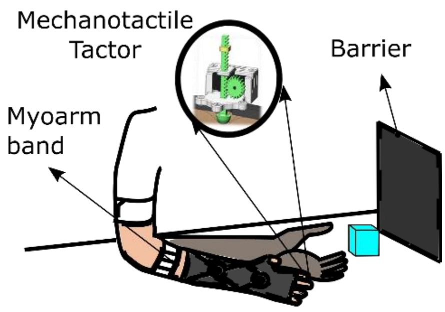
Experimental setup

The experimental setup consisted of a robotic hand with four fingers and a thumb that were driven simultaneously using a linked bar mechanism attached to a single Dynamixel servo motor (MX-64AT). This hand allowed for only 1 degree of freedom for hand open/close. Participants controlled the robotic hand using isometric muscle contractions sensed by an array of eight low power multi-channel operation electrodes (Myoarm^{TM} band) placed around their forearm [18]. A PC running MATLAB (Release 2019b, The MathWorks, Inc., Natick, Massachusetts, United States) and BrachIOplexus software [19] was used to record the control signals during the experiment. The controlled robotic hand was attached to the participant using a modified commercially available wrist brace (MedSpec Ryno Lacer)

Table 1: *Sense of Agency questionnaire items adopted from* [22]. *(A) denotes Agency; (AC) denotes a control question*

Item	**Type**
The prosthetic hand moved just like I wanted it to, like it was obeying my will.	(A)
I felt like I was controlling the movements of the prosthetic hand.	(A)
I felt like I was causing the movement that I saw.	(A)
When I initiated movement, I expected the prosthetic hand to move in the same way that I intended.	(A)
I felt like the prosthetic hand was controlling my will.	(AC)
I felt like the prosthetic hand was controlling my movements.	(AC)
I could sense the movement coming from somewhere between my real hand and the prosthetic hand.	(AC)
It seemed like the prosthetic hand had a will of its own.	(AC)

a)



b)

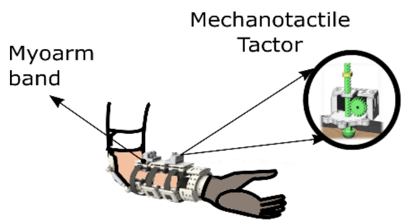


Figure 1: *Experimental setup. a) An able-bodied participant wearing the simulated prosthesis with mechanotactile tactors attached to their index and thumb fingers. Note that the participant's hand is covered by a black sheet during the experiment. b) A participant with* a *transradial amputation wearing the modular transradial* prosthesis *with mechanotactile tactors placed on their residual limb.*

that restricts hand and wrist movements. Two mechanotactile tactors [20] were fitted on the participant's index and thumb fingers of the restricted hand and noise-cancelling headphones playing Brownian noise were placed over the participants ears to ensure that audio cues were occluded. A black sheet (1 x 1 m) was placed over the able-bodied participant's shoulder to ensure that their arm was completely obscured, encouraging embodiment of the prosthesis hand (Figure 1. a). The same experimental setup can be used for persons with transradial amputation by replacing the simulated prosthesis system with a modular transradial socket [21] with the mechanotactile tactors placed on the residual limb (Figure 1. b).

Experimental protocol

Participants: 2 able-bodied female participants over the age of 18 years were recruited for this study. Written informed consent according to Rhodes College IRB was obtained from participants before conducting the experiment.

Participants wore a simulated prosthesis (Figure 1. a) and sat comfortably in front of a table that had a cube (57mm x 57 mm x 57 mm) on it and a barrier (W x H: 25 x 14.5 cm) placed perpendicular to the surface of the table. Mechanotactile tactors were placed on their fingertips and electromyography (EMG) signals from the wrist

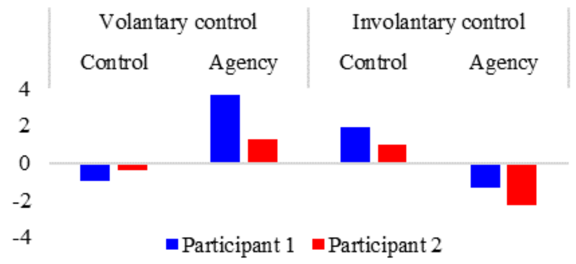


Figure 2: *Average response to sense of agency questionnaire items for voluntary and involuntary control*

flexor/extensor muscles were mapped to the prosthetic hand's open and close controls. The contact forces on the prosthetic hand were mapped to the function of the mechanotactile tactors placed on the participant's fingers. The experimental protocol consisted of the following 3 blocks.

Block 1: Voluntary control

Participants controlled the prosthesis using the calibrated EMG controller and received tactor feedback on their fingertips when the prosthetic hand contacted an object, with a force proportional to that which was placed on the prosthesis' fingers. Each participant was asked to operate the prosthesis to complete a grasp-and-release task that consisted of grasping an object, transporting it over a barrier, placing it down, and releasing the object (30 trials with a 2-minute break halfway through). After this training, the participant was asked to fill out an embodiment questionnaire. Table 1 shows a list of the sense of agency items that were on that embodiment questionnaire.

Block 2: Involuntary control

The experimenter controlled the opening and closing of the prosthetic hand and the participant received tactor feedback on their fingertips when the prosthetic hand grasped the object. The participant was asked to mimic the prosthetic hand movement by contracting the muscles corresponding to this observed movement during the grasp and release phases of the task [grasping the object, transporting it over a barrier, placing it down, and releasing it (30 trials with a 2-minute break halfway through)]. After these trials, the participant filled out the embodiment questionnaire.

Block 3: IB familiarization and testing

During the familiarization phase, the participant was asked to grasp an object, and attend to the moment when the prosthetic hand began to move and the moment that they received the tactor feedback. Ten trials of voluntary control familiarization occurred before the ten trials of involuntary control familiarization. The testing phase of this block included pairs of trials; one voluntary trial and one involuntary trial. Involuntary trials included variable speeds (either faster or slower control). Participants were presented with the trial pairs and were asked to indicate which of the two trials felt faster. If participants were correct, the difference between the speeds of the two trials was reduced until the participant achieved a 50% correct response rate. If they were incorrect, the difference between the two trials was increased (following an adaptive staircase method). The test progressed until the termination condition of the adaptive staircase was reached (23 reversals). The final value achieved indicated the participant's perceived action-effect intervals for voluntary control with respect to their perceived action-effect intervals for involuntary control.

Outcome measures: Data included the responses to the embodiment questionnaire items, rated on a visual analogue scale (0-10). This questionnaire consisted of 20 items that were randomly ordered. In this paper, we focus on 8 of these items pertaining to the sense of agency. The mean of the responses to the 4 agency items for each participant and the mean of the responses to the 4 agency control items were reported. Data from testing block 3 are not reported here.

RESULTS

Similar to our previous study [23], the average responses to agency questionnaire items for the voluntary control with mechanotactile feedback were at least 4.2 times higher than the average responses to control agency questionnaire items. Conversely, the average responses to agency questionnaire items for the involuntary control with mechanotactile feedback were at least 1.6 times lower than the average responses to control agency questionnaire items. These results indicate that the sense of agency as measured using a subjective questionnaire may be affected by voluntary and involuntary control conditions. Comparing participants' average responses to agency questionnaire items between voluntary and involuntary conditions show that involuntary control may have a negative effect on sense of agency and, therefore, the overall embodiment of a device.

DISCUSSION

The aim of this study was to determine the influence of voluntary and involuntary control on the sense of agency with a commonly used questionnaire assessment [22]. This step is crucial for the development of an experimental protocol to objectively quantify the sense of agency by correlating it with intentional binding. We found that involuntary control with feedback may reduce the sense of agency as determined by the administered questionnaire. This finding may be a result of participants not being in control of the prosthetic hand, but also could have been driven by any unexpected effect of the prosthesis touching an object. In our recent work [23], we showed that even with voluntary control, delaying the mechanotactile feedback (> 500 ms) can negatively influence responses to agency questionnaire items. It is worth noting that responses to control questions were slightly affected by the order of condition presentation. These observations warrant an investigation of an objective assessment procedure that allows for an unbiased approach and simple reporting method. We propose to utilize IB and the agency questionnaire, to further investigate the roles that IB, agency, and embodiment play in advanced myoelectric prosthesis use. This investigation can be implemented in able-bodied participants with the use of a simulated prosthesis, or in participants with amputation, which will allow for the investigation into IB in naïve as well as experienced myoelectric users. The methodology presented will allow for quantification of IB in a range of prosthesis users with various sensory feedback strategies in a standardized manner.

A standardized IB method will allow for more efficient data comparison between research centres. To evaluate this assumption, we plan to implement this protocol in a multi-site investigation with a collaboration between three research centres at the University of Alberta, Edmonton, AB; the University of New Brunswick, Fredericton, New Brunswick; and Rhodes College, Memphis, TN, USA. The breadth of this investigation will assist in moving the field of embodiment research toward a more standardized approach, especially for the investigation of psychophysical phenomenon in myoelectric prosthesis users. A standardized methodology will lead to more efficient evaluation of myoelectric devices and technology, prosthesis training protocols, and evaluation of prosthesis embodiment.

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Track: Prosthetic Devices and Materials

ABSTRACT MYOELECTRIC CONTROL WITH AN ARDUINO-BASED SYSTEM

Hancong Wu, Sigrid Dupan and Kianoush Nazarpour

Intelligent Sensing Laboratory, School of Engineering, Newcastle University, United Kingdom

ABSTRACT

This paper presents the design and evaluation of an Arduino-based system for electromyogram (EMG) signal measurement and prosthesis control with the abstract decoder. It achieves a 2 kHz sampling rate for two EMG channels, processes EMG signals on-the-fly and sends the prosthesis control command via a CAN bus. We tested the accuracy and responsiveness of the system in real-time by playing back previously recorded EMG signals through a Tip, Ring, and Sleeve (TRS) function generator. The correlation coefficients between the mean absolute value (MAV) of the original signals and the measured signals were above 97%.

INTRODUCTION

In clinical settings, myoelectric control is achieved by dual-site bang-bang control. Other methods such as pattern recognition, direct control, regression and abstract decoding have been introduced as alternatives [1]. Pattern recognition extract features from the EMG signals and groups the inputs into discrete movement classes. This technique often entails complex machine learning procedures, and it is normally implemented on high-performance processors [2, 3]. Recently, customised embedded electronic systems have been developed to enable real-time prosthesis control to approximate clinical settings [4, 5]. However, the width of adoption of the embedded system, as a research tool, is slow due to the cost and resources that are required to develop a reliable, flexible system.

In this work, we introduce a simple Arduino UNO-based embedded sensing and processing system for prosthesis control with abstract decoding [6, 7]. We evaluate the function and reliability of the system using previously-recorded EMG signals.

METHODS

System architecture

The conceptual design of the proposed embedded system is presented in Figure 1.

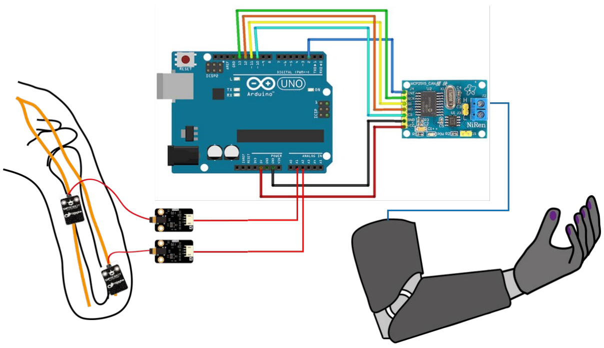


Figure 1: Conceptual design of the proposed embedded system

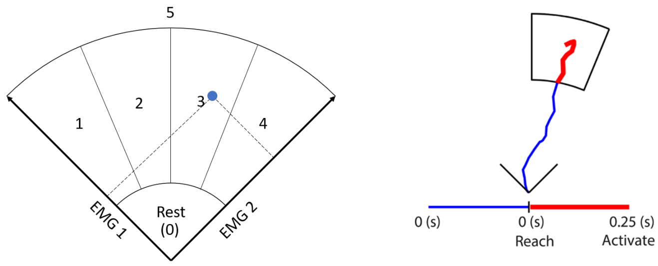
Our Arduino UNO-based system comprises four modules for data collection, EMG signal processing, prosthesis control and data transmission. The data collection module can sample up to two channels of EMG as fast as 2 kHz per channel. The signal processing module works at 100 Hz. It removes the DC bias of the input data, reduces the signal noise through an averaging filter and normalises the EMG signal based on calibration, in accordance with Dyson et al. [8]. The control module determines the grip patterns with an abstract decoder and sends the motor commands to the robo-limb™ prosthetic hand (Össur, Reykjavík, Iceland) via the data transmission module.

Abstract decoder

Unlike machine learning-based approaches, abstract control relies on human learning for the operation of the myoelectric-controlled interface (MCI) [6]. Abstract decoding promotes the co-contraction of muscle groups that are not co-contracted naturally for new functional gains or the utilisation of natural co-contractions. An example of the MCI is as outlined in Figure 2 (a). In our proof-of-concept implementation, we split the control interface into six zones, named the rest zone (zero), grip zones one to four and the outlier zone (five). Users control the instantaneous position of the blue 2D cursor with the control signals that are extracted from the two EMG signals. To activate a grip on the prosthesis, the user should hold the cursor in a grip zone (one to four) for a certain period. Figure 2 (b) shows a representative cursor trajectory for an individual trial.

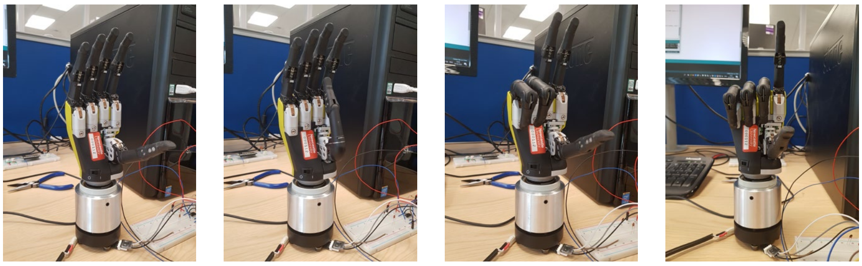
In our implementation, the cursor timer goes to sleep when the cursor stays at the rest zone or the outlier zone. Once the cursor moves to a grip zone, the timer records the period when the cursor is held within it. A grip command associated with the zone is sent to the prosthesis if the cursor stays within the zone for $t = 0.25$ seconds. The movement of the cursor to another zone will reset the timer (base time: 10 milliseconds). The system will not send motor commands to the prosthesis when the cursor stays at the rest zone or the outlier zone so the hand will maintain at the last grip until a new grip is determined.

In this implementation, we considered four grips, the normal grip, the thumb park grip, the three-jaw chuck grip and the pinch grip, and assigned them to zone one to four, respectively (Figure 3).



(a) (b)

Figure 2: The 2D MCI space and a representative cursor trajectory.



(a) (b) (c) (d)

Figure 3: The grips correspond to (a) zone one, (b) zone two, (c) zone three and (d) zone four

RESULTS

We tested the performance of the proposed embedded system. A MATLAB program controlled the stimulation of the signals through the TRJ function generator, as demonstrated in Figure 4. A potential divider and an amplifier circuit processed the signal to mimic the EMG signal measured by the Gravity Analog EMG sensors (OYMotion Technologies, Shanghai, China). The results are presented in two sub-sections. The performances of the data collection module and the signal processing module are in Section one. Section two presents the functional test of the control module and the data transmission module.

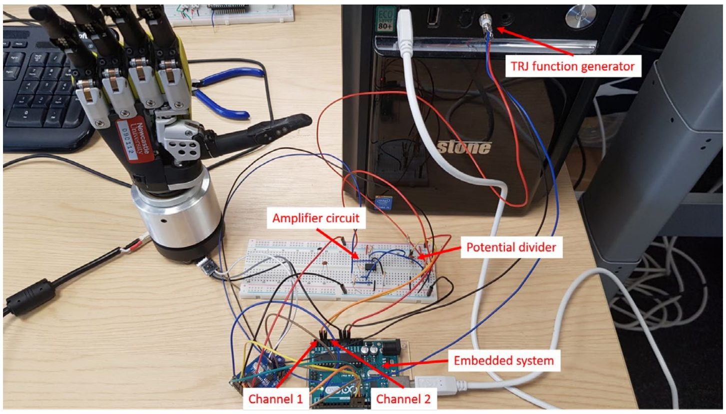
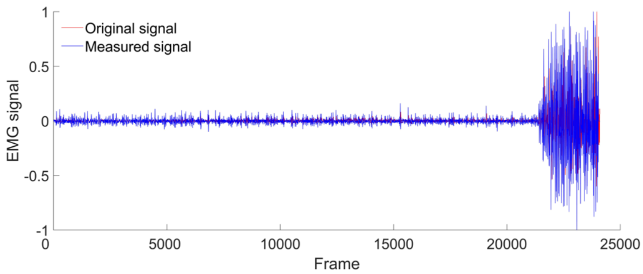


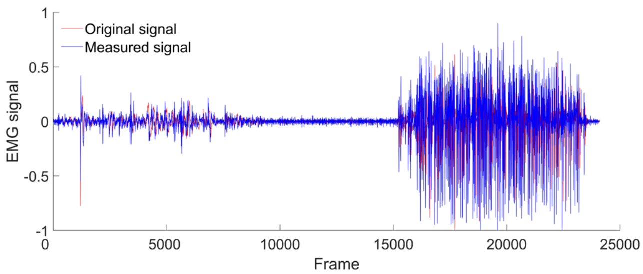
Figure 4: The embedded system connected with a prosthetic hand and TRJ function generator

Analysis of the EMG signal

Figure 5 shows the comparison between the original EMG signals and the signals measured from the embedded system. The original signals were previously recorded by D360 amplifier (D360, Digitimer, UK) at 2 kHz sampling rate for 12 seconds. Since the sampling rate of the embedded system was set to 1 kHz, linear interpolation was applied to the sampled signals to maintain the same length as the original signals. The measured signals closely matched to the original signal. The correlation coefficient between the moving MAV of the original signals and that of the measured signals are 99.43% and 97.58% at two channels.



(a)



(b)

Figure 5: The comparison between the original EMG signals and the measured signals at (a) channel one and (b) channel two.

Evaluation of the control module

The state of the prosthesis controller is changed by the control signals. Figure 6 shows an example in which the embedded system sent two motor commands to the prosthetic hand. The first command was sent at the 856^{th} frame, which was 0.75 second (75 frames) after the increase in the control signal on channel two. It changed the prosthetic hand from the normal grip to the pinch grip. The second command was sent at the 1279^{th} frame, which was 0.37

second (37 frames) after the participant released the muscle on channel 1. The prosthetic hand returned to the normal grip after receiving the command. The time required to change the state of the prosthesis was keeping the cursor position at the same zone for 0.25 second (25 frames) as expected. Although the cursor temporally moved to zone two between the 1243^{th} frame and the 1254^{th} frame, the abstract decoder did not send a command to the prosthetic hand.

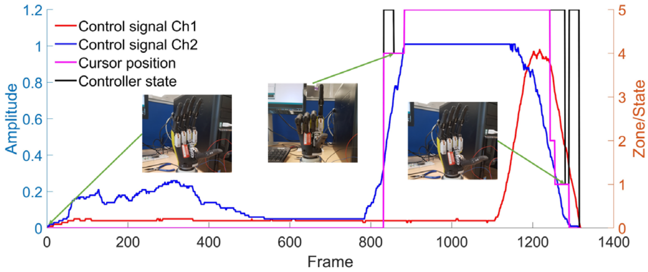


Figure 6: The EMG control signals and the corresponding changes in the state of the prosthesis

DISCUSSION

The analysis of the EMG signals and the controller states demonstrates that the Arduino development board is capable of EMG data collection. The measured signals on both channels maintain high similarity with the original signals generated from the TRJ function generator. The abstract decoder working at 100 Hz can correctly indicate the location of the cursor and control the prosthetic hand with a 10-millisecond temporal resolution. Its simplicity and low computational cost requirement allow it to be implemented on the Arduino board.

This paper presents a new embedded system with off-the-shelf components that allows myoelectric control through the abstract decoder. With a £17 equipment cost, the proposed system can achieve a maximum 2 kHz sampling rate for 2-channel EMG measurement and the real-time prosthesis control. It removes the barrier for many researchers to perform take-home experiments without designing a customised embedded system. We aim to present a demo of this system at MEC2020 and release the design specifications and code in due course.

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BASIC EXPERIMENT TO COMPARE THE PERFORMANCE OF MULTI-ARTICULATED 3-DIGIT MYOELECTRIC HAND

Kengo Ohnishi1, Tomohisa Morita1, Gai Higuchi1, Hibiki Takami1, Daisuke Kuwayama2, and Kazuhiko Urata2

1: Tokyo Denki University, 2: P.O.care center Tetsudoukousaikai Fundation

ABSTRACT

The anthropomorphic hands with multi-articulated digit are appealing compared to conventional 1-Degree-of-Freedom hands with 3-digits 3-joints. The coordinated movable joints ease to grasp objects with variety of shape and size while reducing compensative joint movements of the residual limb and torso. Therefore, the developers state their design superiority based on the number of capable prehensile forms, which are important index to describe the static function of holding object(s) within the hand. However, to conduct tasks, the response and efficiency grasping motion are nevertheless important. Further understanding the effect of multi-articulated digit design on the static and dynamic functions, in relation with myoelectric control is a notable topic. In this research, we conducted a comparative experiment between 3-digit hands: 3-joint conventional, Ottobock DMC hand, and 7-joint multi-articulated fingered THK TRX hand. Gripping time of pick-and-place task on large and small diameter cylinders were measured. 3 non-amputee subjects participated in the test using hand mounted on a quasi-prosthesis socket. As result, the large diameter cylinder's average gripping time of 7-joint hand was 0.77 seconds, larger than that of the 3-joint hand, 0.42 seconds. For the small diameter cylinder, the gripping time for 7-joint hand was shorter than the 3-joint hand.

INTRODUCTION

The anthropomorphic prosthetic hands with multi-articulated digit are appealing compared to conventional 1-Degree-of-Freedom (DoF) hands with 3-digits 3-joints structure. The coordinated movable joints of anthropomorphic hands ease to grasp objects with variety of shape and size while reducing compensative joint movements of the residual limb and torso. Therefore, the developers state their design superiority based on the number of capable prehensile forms [1, 2], which are important index to describe the static function of holding object(s) within the hand. However, to conduct manipulation tasks, the dynamic functions, i.e. response and efficiency gripping motion, are nevertheless important. Fukuda et al. [3] reports difference between the proportional speed and the constant speed myoelectric control of a hand on the screen. The trajectory of approaching the hand to the target and the gripping time operating the virtual hand on a monitor showed different timing of closing the hand. Bouwsema et al. [4] reports on their investigation of two grasping tasks and a reciprocal pointing task of a myoelectric transradial prosthesis in comparison to intact hand. Decoupling of reach and grasp were reported with other characteristic of kinematics of grasping. Experiment to analyze the trajectory and motion time of conducting tasks with myoelectric hand with multi-articulated digit should indicate design solution to improve static and dynamic functions of the hand mechanism. In this paper, a basic experiment is reported comparing the performance of 3-digit myoelectric hand with multi-articulated and traditional digits.

METHODS

Two 3-digit 1-DoF hand is compared. As a conventional prosthetic hand, Ottobock 8E38=6 DMC plus (referred to as DMC hand) is applied. The MP joint in the digits are coupled and driven. As Multi-articulated digit hand, THK TRX hand prototype (referred to as TRX hand) is applied. The index and middle fingers are designed with link-coupled 3-joint mechanism. The PIP joint of the finger flexes in relation to the MP joint, and a coil spring in the DIP joint operates to adapt the joint angle to settle the contact. The thumb's MP joint is driven in relation to the finger's

MP joint with a linear actuator. The TRX hand is operated by two microcomputers: THK SEED-MS3A (referred to as MS3A) and SEED-BL1A (referred to as BL1A), as in Figure 2. The MS3A acquires and process the myoelectric sensor signals and BL1A operates the linear actuator speed for opening and closing the hand.

To experiment the effect of articulated finger on the reach-gasp-pick-place-release task, subjects' operations of the hands were measured and the gripping time are compared. Motion capture and analysis system (Nobby Tech, VENUS 3D), with 4 cameras (Optitrack Flex 13, 1280x1024, Sampling:120 Hz) are arranged on the desk and 6 markers (Diameter:6.6 mm) were attached to the hands in relation to compute the joint angles of the digits and hand position and posture (Figure 3). The experiment was approved by the TDU IRB (#28-97, 29-83, 30-65, 31-094). Three able-bodied subjects (average 23.3 SD0.4 years-of-age, all male and right handed) participated donning a quasi-transradial socket with hands connected distal to the sound hand, after giving a written consent (Figure 3). Two objects, a large diameter cylinder (D:48 mm, H:100 mm, 112 g) and a small diameter cylinder (D:30 mm, H:100 mm, 42 g), were to be grasped at the lateral side. The large diameter cylinder was requested to hold with power grip form, and small diameter cylinder with precision grip form at the distal phalanges. The right direction was set as positive of X axis, the depth direction as positive Z axis, and the vertical up direction as Y axis (Figure 4). The initial position of the hand was set on the edge of the desk. The object initial position was set on the desk (H:780 mm) at the position Z:300 mm, X:-100 mm and released at position Z:300 mm X:100 mm. The subject was seated at H:440 mm, Z:-300m, in front of the desktop workspace. The experimental was proceeded as follows:
1) Adjust the width between the thumb and finger to 50 mm by grasping a wooden rectangular parallelepiped (50x50x100mm) before measurement.
2) Place the hand in the initial position.
3) Reach, gasp, pick, move the object to the right 200 mm and release.
4) Returned the hand to the initial position.
The experiment was conducted with the small diameter cylinder first and the large diameter cylinder latter for both hands. Subjects practiced 5 times before measurement. Total of 120 trials (10 trials for 2-objects, 2-hands, 3-subjects) were recorded and the latter 5 trials in each condition were used for evaluation.

Figure 1: DMC (top) and TRX (bottom) hand's full open/close state

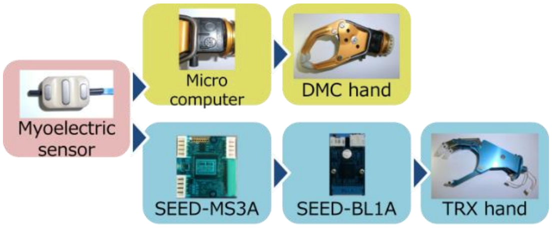

Figure 2: DMC hand and TRX hand's controller components

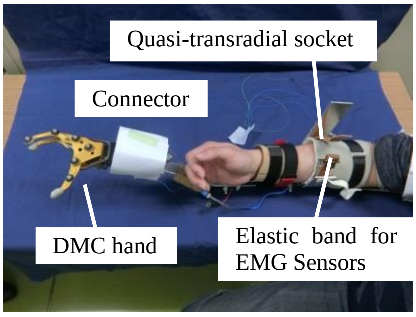

Figure 3: Experimental setup of the DMC hand on quasi-transradial socket

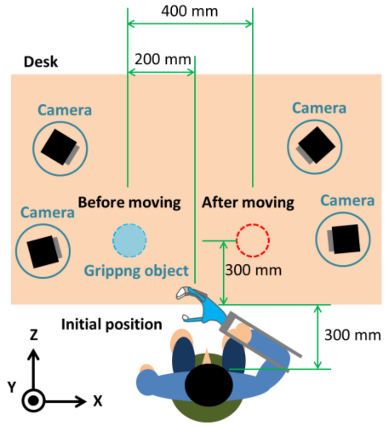

Figure 4: Experimental setup for measurement of pick-and-place task of small and large diameter cylinder.

RESULT & DISCUSSION

The gripping time, which consist of time during the reach and grasp motion, of the hands were calculated from the joint angle valuation of the index finger. The results showed that 3-joint DMC hand took approximately twice longer time to grip the small diameter cylinder, average 0.73 s, than the 7-joint TRX hand, 0.38 s, and on the contrary, DMC hand took approximately half to grip the large diameter cylinder, 0.42 s, than the TRX hand, 0.77 s (Figure 5). T-test showed that the differences were significant: the small diameter cylinder ($t=2.22$, $df=28$, $p<0.05$), and the large diameter cylinder ($t=-2.38$, $df=28$, $p<0.05$).

The changer over between the cylinder diameter size is further inspected. Light et al. reported, that the mean norm task time of power grasping objects, heavy- and light-weight, is longer than that of precision (tip and tripod) grip in their result analysis experiment with the SHAP test with intact upper limb subjects [5]. The 7-joint hand shows a resembling result to the SHAP test, however the time for small diameter cylinder is shorter. This is due to the initial “hand open” finger position to make the fingertip travel smaller. The articulated mechanism allows the fingertip to be moved faster making the 3-joint hand slower to grip. On the large diameter cylinder, the 3-joint hand became faster with the hand gripping the object with the proximal part of the hand. The assumption of the cause of this was that digit shape of the 3-joint hand makes the Form Closure of the grip [6] easier and with less movement to enable short gripping time. To confirm this assumption, the time to the contact of the cylinder and the hand were computed (Figure 6). The 7-joint hand contact was 0.17 s and shorter than that of 3-joint hand. The T-test between the hands showed the difference was significant ($t = 2.68$, $df = 15$, $p > 0.05$). This showed that 7-joint hand required additional time to have the distal phalanges to contact the object for Form Closure to stabilize the force to proceed to pick the object. The time surrounded by the dotted line on the bar of TRX_large can be considered as the drawback of mimetic mechanism of the human digit. Furthermore, this lag can be the disappointment of the DMC hand user to feel that multi-articulated hands to be cumbersome.

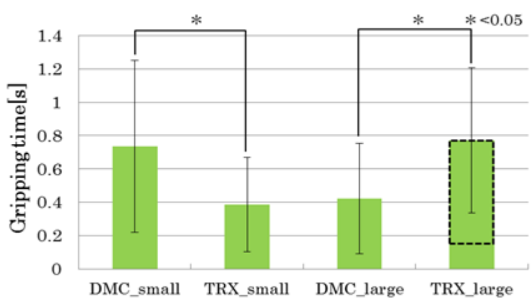

Figure 5. The gripping time in each condition

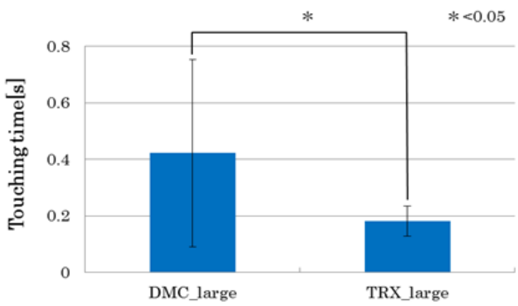

Figure 6. Average time till contacting the object with the hand

CONCLUSION

The effect of the multi-articulated finger control to perform pick-and-place task was discussed by comparing the gripping time of small and large diameter cylinders. A 7-joint THK TRX hand and the 3-joint Ottobock DMC hand are operated by non-amputee subject and motion capture system is used to evaluate and compare the result. The results showed that 3-joint DMC hand took approximately twice longer time to grip the small diameter cylinder, average 0.73 s, than the 7-joint TRX hand, 0.38 s, and on the contrary, DMC hand took approximately half to grip the large diameter cylinder, 0.42 s, than the TRX hand, 0.77 s. Further inspection described that the time for Form Closure of the 7-joint hand causes addition time to power grasp compared to the 3-joint hand. Further analysis of the motion captured data and modified experimental design is required to investigate the design factors of prosthetic hand mechanism and myoelectric control and their ponderability to the performance of object manipulation.

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DEVELOPMENT AND EVALUATION OF POINTDEXTER – AN INTEGRATED PREHENSOR FOR PROSTHETIC FINGERS

Carlos Martinez-Luna[1]; Benjamin Bokser[1], Carlton King[1], Todd Roberts[1], Brianna Rozell[1], Todd Farrell, Ph.D.[1]

[1]Liberating Technologies, Inc. – A College Park Company

ABSTRACT

Current prosthetic terminal devices require a compromise between form and function. Pointdexter is a retrofittable miniature gripper that is integrated into the index finger of multi-articulating hands to allow for an additional, selectable, grasp to assist in the manipulation of small objects. Pointdexter is an all-mechanical design that does not require additional actuators and is controlled using existing prosthesis control signals. Testing on able-bodied and amputee test subjects was performed using the Jebsen Taylor Hand Function test using three terminal devices: an unmodified Bebionic hand, the Bebionic with Pointdexter, and a Motion Control ETD. The results demonstrate that Pointdexter improved small object manipulation time over an unmodified multi-articulating hand by >35%, while not impacting normal hand function. Additionally, take home testing was performed to identify additional areas of improvement and to evaluate robustness of the device.

INTRODUCTION AND BACKGROUND

The Problem

Current prosthetic terminal devices (TDs) each have their advantages and disadvantages, which requires a compromise between form and function (Figure 1). Some individuals will carry multiple TDs and swap them out based on the environment and task being performed.

The Solution

A dexterous fingertip terminal device, Pointdexter, (Figure 2) was designed to optimally combine the advantages of multi-articulating prosthetic hands (e.g., conformal grasp) and hooks/grippers (e.g., small object manipulation) in a single upper-limb terminal device. Pointdexter adds function within the form and aesthetics of multi-articulating hands, as appearance is often as important as function in adoption of the prosthesis by the user [1].

Pointdexter adds an additional, selectable, dexterous grasp option focused on manipulating small objects. In this approach, the pointer finger on the hand is replaced with the self-contained and retrofittable Pointdexter to provide a tonged end-effector at the fingertip. The current, all-mechanical design is an add-on to existing multi-articulating hands that does not require additional actuators.

Pointdexter is driven with standard control signals. It is activated during 'trigger' grip via a selectable mechanical mode switch (Figure 3). When the Pointdexter is locked, the jaws are closed and finger is free to flex and extend as it normally would. When Pointdexter is unlocked, the finger actuator opens and closes the tines of the gripper instead of flexing and extending the finger.

After development of the prototype, functional outcomes measures were used to quantify the change in small object grasping ability created by Pointdexter and also confirm that Pointdexter does not interfere with standard hand function.

	Grasps Large Objects	Grasps Small Objects	Conformal Grasp	Multiple Grip Patterns	Cosmetic
Split Hook	✓	✓	✗	✗	✗
Electric Gripper	✓	✓	✗	✗	✗
Conventional Electric Hand	✓	✗	✗	✗	✓
Multi-Articulating Hand	✓	✗	✓	✓	✓
Multi-Articulating Hand + Pointdexter	✓	✓	✓	✓	✓

Figure 1: Commercially available TDs (top) and a feature comparison matrix (bottom).

Figure 2: Pointdexter features.

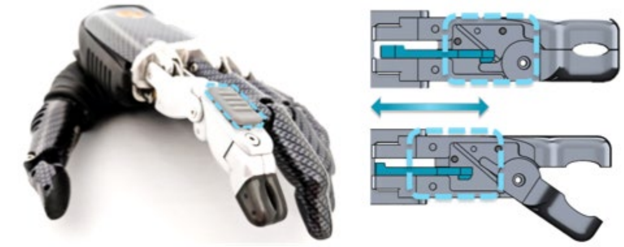

Figure 3: The 'top lock' mode switching mechanism.

METHODS

LTI conducted an initial clinical evaluation of functional outcome measures to compare Pointdexter (Liberating Technologies), a Bebionic hand (Otto Bock), and a powered split-hook ETD (Motion Control) (Figure 4). The ETD was selected to serve as the 'gold-standard' for function. IRB approval and participant informed consent was obtained.

A first round of testing for protocol development included subjects conducting three repetitions of the Jebsen-Taylor Hand Function (JTHF) test, 9-hole peg test, and common bimanual tasks. However, during this testing it was discovered that fatigue was substantial and was likely affecting the results. Therefore, the second round of testing that is described here focused solely on performing three repetitions of the JTHF test.

Figure 4: The 3 terminal devices used for testing: standard Bebionic hand (left), Bebionic with Pointdexter (center), and ETD (right).

<u>Subjects</u>

Two persons with transradial limb absence and two able-bodied subjects using prosthesis simulators (Figure 5) have participated in this study to date. Amputee subjects were experienced (>6 months) users of myoelectrically controlled multi-articulating hands.

Figure 5:A photograph of the able-bodied simulator.

<u>Procedures</u>

Participants practiced with each device to reduce the potential for learning effects. Subjects then conducted three timed trials of each sub-task of the Jebsen-Taylor with each of the three terminal devices. The terminal devices were presented in random order. Before the start of each task, subjects were allowed to select the desired grasp pattern in the standard Bebionic condition. During the Pointdexter condition, the subjects also had the option of selecting to use the Pointdexter or not. The selection of whether or not to use Pointdexter was consistent across all subjects. Every subject chose to use Pointdexter for turning cards, lifting light cans, stacking checkers, and manipulating small objects, but not use it for simulated feeding, writing, and lifting heavy cans (Figure 6).

<u>Data Analysis</u>

Mean and variance data of the able-bodied and amputee subjects were similar, so the results were pooled. Full statistical analyses were not conducted due to the small number of patients in this pilot study.

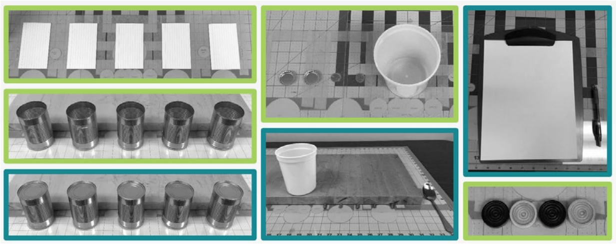

Figure 6: The subtasks of the Jebsen-Taylor test. When allowed, subjects chose to use the Pointdexter for those in green but not for those in teal.

RESULTS

Figure 2 shows the average completion times and 95% confidence intervals across test subjects for the Jebsen-Taylor Small Common Objects functional task conducted with each device. The standard multi-articulating hand was the slowest and the ETD the fastest, with the Pointdexter being >30 seconds faster (a >35% improvement) than the unmodified hand. As expected, both hand conditions were slower than the ETD on the small objects task.

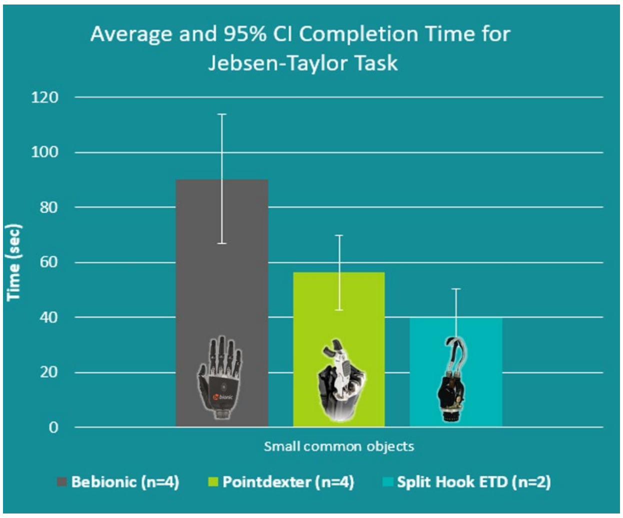

Figure 7: Jebsen-Taylor small objects test completion times for the tested terminal devices. Average across subjects with 95% confidence intervals.

Performance was found to be similar between the Pointdexter and the unmodified Bebionic hand for all other subtasks of the JTHF except card turning. More detail on this is provided below.

DISCUSSION

The Pointdexter design aims to combine the best aspects of various terminal devices and eliminate the need for users to frequently physically transition between terminal devices to accomplish various tasks/ADLs requiring dexterity.

As expected, the powered split-hook ETD performed the best across tasks and users as it is generally considered the most functional device we tested. However, the Pointdexter was able to emulate the precision of the split-hook ETD in manipulating small objects and improve the performance when compared to the standard multi-articulating hand. The variability was high and the sample size too small for statistical analysis, so further testing is required.

It was interesting the see how many tasks on the Jebsen-Taylor test individuals voluntarily chose to use Pointdexter. We believe that this has to do with the novelty of a new device or feature. For example, all subjects chose to try to pick up the large empty cans with the Pointdexter. However, we believe that, after using the device outside of the lab, it is likely that subjects would not choose Pointdexter for this task. Similarly, we believe that the increased time to complete the card turning task was due to the fact that the gripper on the Pointdexter was fairly small and therefore required more precise alignment to grip the card, while in full-hand mode there is a larger width of opening and therefore a larger margin for error. We believe that real-world practice would identify which tasks are best suited for Pointdexter and optimize its usage.

TAKE-HOME TRIAL

In addition to the in-lab testing described above, we conducted a one-month take home trial to identify:

- Areas of Improvement – more grip strength was the primary request.
- Tasks it was particularly useful for (Figure 8).
- Potential robustness issues – fortunately there were none.

Figure 8: A photograph of in-home use of the Pointdexter.

CONCLUSION

Initial functional tests with the Pointdexter are encouraging. Adding Pointdexter to a multi-articulating hand improved the user's ability to grasp small objects while retaining normal hand function and anthromorphic shape of the hand. Ideally, this design will increase prosthesis use and thus help to decrease overuse injuries in the intact limb from the relatively young UL amputee population.

ONGOING / FUTURE WORK

Additional research funding has been acquired to continue the project and implement various design changes and expand functional testing with human subjects. Anecdotal feedback from users highlighted a desire for more precise, secure, and strong grip patterns in the multi-articulating hand. Design efforts are underway to improve strength and security of grasp in order to gain even more functionality. Several changes have been implemented and initial functional tests with the improved design are encouraging. Also, while the Pointdexter was originally designed to work with the Bebionic hand, a new version has been developed to integrate with another popular multi-articulating hand, the iLimb from Össur (Figure 9).

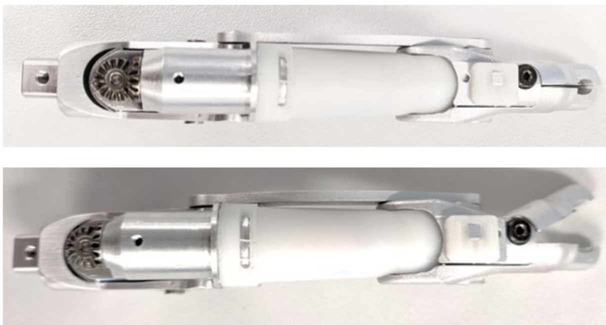

Figure 9: Photographs of the Pointdexter designed to integrate with the iLimb hand from Össur.

ACKNOWLEDGEMENTS

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DEVELOPMENT OF A MODULAR SIMULATED PROSTHESIS AND EVALUATION OF A COMPLIANT GRIP FORCE SENSOR

Eric D. Wells [1], Shealynn Carpenter [2], Michael R. Dawson [3], Ahmed W. Shehata [3], Jason P. Carey [1], Jacqueline S. Hebert [3, 4]

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

ABSTRACT

Grip force sensory feedback is commonly stated as a desirable feature for upper-limb myoelectric prosthetics. Many techniques for non-invasive grip force feedback are being investigated. However, the choice of force sensor, feedback location, and experimental apparatus typically vary between research studies, making it challenging to compare results. A standardized device where individual parameters can be adjusted would allow researchers to evaluate the impact of each variable on results. An example of such a device is a simulated prosthesis. Simulated prosthesis devices enable non-disabled individuals to participate in myoelectric prosthesis research experiments while ensuring consistency in experimental apparatus between participants. We developed a lightweight, modular, and inexpensive simulated myoelectric prosthesis capable of delivering sensory feedback to fingertips and proximal forearm. We integrated mechanotactile feedback devices to deliver modality matched feedback to the forearm and somatotopically matched feedback to the fingertips. We compared a commercial force sensor before and after being encapsulated within a compliant material under a variety of loading conditions. The encapsulated force sensor outperformed the standard sensor in all non-ideal loading conditions by a large margin. The use of this encapsulation technique dramatically increases accuracy in sensor readings when loading conditions differ from calibration conditions. This device will help facilitate myoelectric research by providing a consistent experimental apparatus between non-disabled participants for various control and feedback-oriented studies.

INTRODUCTION

Upper limb amputation results in loss of both motor and sensory function of the hand, harming an individual's economic, psychological, and social well-being [1]. Prosthetic technology attempts to mitigate these effects by restoring functionality to the lost limb. Current research in the upper limb prostheses field focuses on electrically powered devices controlled by the muscle signals in the residual limb, termed myoelectric prostheses [2]. Myoelectric devices utilize the existing neural pathways in an open-loop fashion, without specific feedback on the outcome of the action.

Upper limb myoelectric prostheses users commonly state sensory feedback as a desirable feature, with grip force ranking as the highest priority sensory input [3]. Many methods of non-invasive grip force feedback implementation are being investigated with promising results [4]. However, parameters such as feedback location, force sensors, and experimental apparatus are typically unique to each experiment, making comparisons between studies difficult. There is an ongoing need for devices capable of adjusting these parameters to allow researchers to evaluate each variable independently.

In previous studies, simulated prosthesis devices have been used to investigate myoelectric control [5] and sensory feedback techniques [6]. An evaluation of a simulated prosthesis device showed that it resulted in motion kinematics and performance metrics similar to those found in myoelectric users [7]. A Simulated Sensory Motor Prosthesis previously constructed within our lab allowed for somatotopically matched mechanotactile feedback during myoelectric control [8]. However, initial testing with the device showed various issues that justified a revision. The large size, non-modularity and weight of the device (1.3 kg) made it difficult to move naturally, causing discomfort over long periods.

The objective of this work was to optimize the size, weight, and comfort of the Simulated Sensory-Motor Prosthesis while maintaining the ability to provide sensory feedback to both the forearm and fingertips. This allows for both modality and somatotopically matched feedback to be used on the same experimental apparatus. An additional focus was placed on modularity to allow for interchangeable components for various user sizes or experimental conditions. The device was fit with inexpensive compliant force sensors to measure the grip force of the end effector reliably. These sensors were evaluated and compared to standard sensors under various loading conditions to ensure accurate grip force measurement.

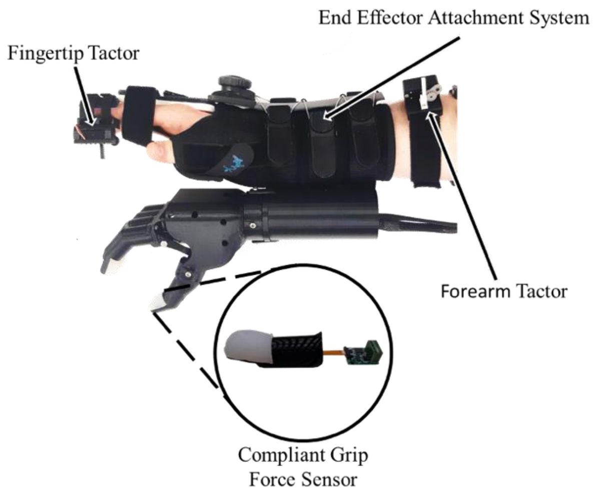


Figure 1: The MSP Overview

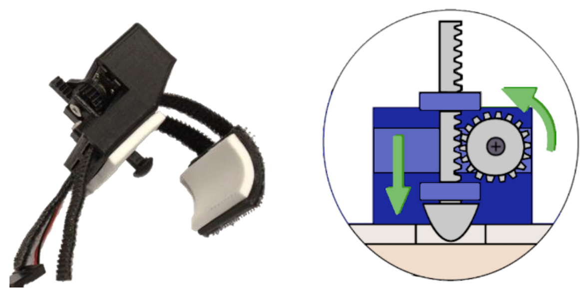
MECHANICAL DESIGN

Figure 1 shows an overview of the Modular Simulated Prosthesis (MSP) that was developed. A wrist and thumb support brace (MedSpec, USA) restrains the user's hand to ensure isometric contraction during electromyography (EMG) control. This commercially available product is designed to be comfortable, lightweight, adjustable, and leaves adequate space on the proximal forearm for EMG sensors and other devices. Additional finger flexion restraints were required to prevent the fingertips from colliding with the end effector. This was achieved by extending the existing metal supports within the brace with 3D printed PLA supports.

In previous simulated prosthesis devices, the prosthetic hand is typically mounted with a distal, radial, or ventral offset. Any combination of these offsets places the additional weight of the prosthetic hand off the axis of the user's arm, resulting in an undesired torque. Because the human hand width is much smaller than its length and breadth, this torque is minimized by offsetting in the ventral direction. An adjustable offset in the radial direction was also added to the MSP to resolve any line of sight issues that may arrive for specific tasks. An end effector attachment system was developed to attach the prosthetic hand to the brace while accommodating a variety of arm shapes and sizes. The system consists of a 3D printed bracket that rests midline on the ventral surface of the wrist brace and a cable tightening system (BOA, USA) that rests midline on the dorsal surface of the wrist brace. Attached to the bracket is a 3D printed wrist adapter for end effector mounting. The bracket is temporarily secured to the ventral side of the arm using a large Velcro strip. The cable tightening system is then wrapped around to the dorsal side, where 3D printed quick-connect clips are connected, completing the loop around the arm. The interlocking cable system is tightened to create a snug fit between the end effector and the participant's forearm to minimize the relative movement of the device.

A 3D printed, anthropometric, single-degree-of-freedom end effector was designed (Solidworks, 2018). The hand is driven by a Dynamixel MX-64AT servo motor (Robotis, Inc.). The fingers and thumb are actuated simultaneously using a linked bar mechanism, giving a gripping aperture of 100 mm. This end effector has a mass of 298 grams with a maximum continuous grip force of 11 N. The total mass of the MSP is 691 g with the end effector included, can be comfortably worn for 3 hours, and costs less than $1000 CAD. The end effector, feedback devices, and attachment system are all independent units creating a highly modular design that can be easily customized to fit specific needs.

SENSORY FEEDBACK DESIGN



(a) (b)

Figure 2: Mechanotactile Tactor Overview: (a) Fingertip Mounting System, (b) Motion Illustration

Sensory feedback is integrated into the MSP using small, inexpensive mechanotactile tactors modified from our earlier work [9]. The tactor devices use a lightweight Dymond D47 servo motor (Dymond, USA) with a 3D printed rack and pinion system to apply force to the user. We developed two mounting systems to apply somatotopically accurate feedback to the fingertips, or modality matched feedback to the forearm. The tactors are secured to the user with Velcro straps. Washable foam provides cushioning to prevent irritation to the user. The tactor with the fingertip mounting system is shown in Figure 2. The tactors can provide up to 12 N of force with a throw of 14 mm.

SENSORIZATION DESIGN AND EVALUATION

Measurement of grip force can be done through small force sensors placed on the fingertip of the prosthetic hand. Capacitive force sensors have previously been shown to perform better than commonly used force-sensitive resistors for this application [9]. These sensors are designed to be attached to a flat surface, with the force loading evenly distributed across its surface area. However, prosthetic hands undergo a variety of loading conditions that do not represent this ideal situation. Prosthetic fingertips with barometric pressure sensors embedded in elastomer [10] have previously been shown to provide pressure sensitivity in non-ideal loading conditions. It was hypothesized that encapsulating a capacitive force sensor in a compliant material would disperse the force evenly throughout the sensor, allowing for more robust measurement to various loading conditions.

Methods

A SingleTact S8-10 capacitive based force sensor (SingleTact, USA) was compared before and after being encased in Dragon Skin 10NV, a compliant silicone rubber based material (Smooth-On, USA). The two configurations are shown in Figure 2. A load cell (Omega LCM703 calibrated to a maximum error of 0.1N) was placed in line with an HS-35HD servo motor (Hitec RCD, USA) to apply force to the sensor through a PLA indenter. The load cell was read using Simulink Real-Time (Matlab 2014a) through a National Instruments data acquisition system (NI PCI6259). A force was applied between 0 and 10 N in a sinusoidal pattern for five total periods, similar to earlier work [9]. Loading periods of 0.5, 1, and 5 seconds were tested to account for dynamic loading effects. Each measurement was repeated three times to ensure repeatability between trials, for a total of 9 trials for each condition.

An indenter was made with a circular flat contact surface (10 mm diameter) and covered in a 2 mm thick foam to ensure even force distribution over the entire surface area of the sensor. Loading of this indenter directly aligned with the sensor acted as the ideal condition for both the baseline and the encapsulated configurations. All other conditions were compared to the ideal condition to evaluate the sensor's ability to adapt to various circumstances. An indenter with a 10 mm diameter curvature was tested to represent grasping a curved surface. The indenter position was moved by 4mm in both the proximal and distal directions to evaluate the effect of a non-central loading condition. For only the encapsulated configuration, a centred applied loading condition at a 15-degree angle was also evaluated.

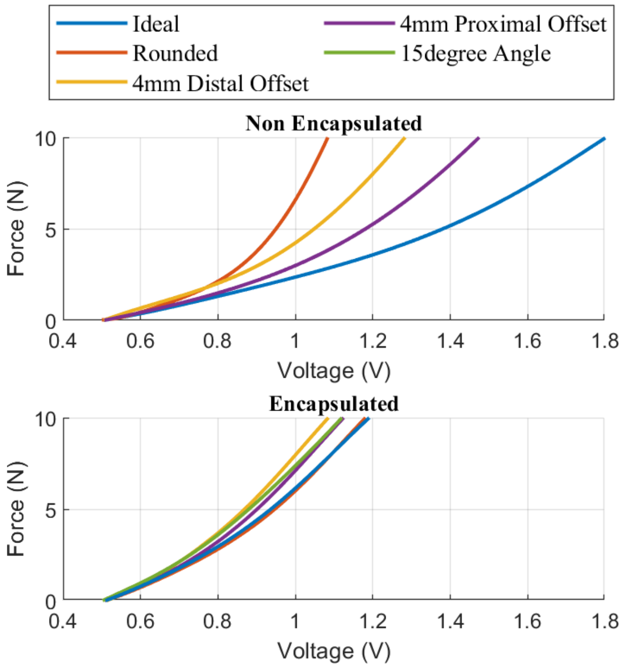


Figure 3: Loading Curve Comparison Between Various Conditions

The baseline and encapsulated sensors voltage to force relationship was calibrated using a 5th-degree polynomial curve fit to all trials under the ideal condition. This calibration curve was used to predict force outputs under all other conditions.

Results

The results for all conditions are summarized in Table 2. In the ideal condition, both sensors performed within the manufacturer's specifications at root mean square error (RMSE) of 2.2% and 2.5% of full-scale range (FS) for the baseline and encapsulated sensor. The RMSE of the baseline sensor was much more sensitive to changing conditions than the encapsulated sensor. The curved indenter condition produced a substantial decrease in performance for the baseline sensor, giving an RMSE of 36.4% FS. The encapsulated sensor was relatively unaffected with an RMSE of 2.9% of FS. Similarly, when the ideal indenter was shifted by 4mm, the RMSE for the baseline rose to 25.5% FS (distal offset) and 15.5% FS (proximal offset). The encapsulated sensor RMSE increased to 10.5% FS (proximal offset) and 7.2% FS (distal offset). Finally, the encapsulated sensor showed an RMSE error of 7.6% FS during the 15-degree angled loading scenario. Figure 3 shows each sensor's loading curve fit with a 5th-degree polynomial curve. The baseline sensor's loading curves are much more varied when contrasted with the encapsulated sensor, illustrating the dependency on environmental conditions. For example, at a load of 10 N, the baseline sensor voltage output varies by 0.72 V (50.7% FS over 10 N) depending on the condition, while the encapsulated sensor only varies by 0.11 V (14.4% FS over 10 N).

Table 1: Summary of Experimental Results for Grip Force Sensor Comparison

Loading Condition	Baseline Sensor RMSE (N)	Encapsulated Sensor RMSE (N)
Ideal	0.22	0.25
Rounded	3.64	0.29
4 mm Distal Offset	2.55	1.05
4 mm Proximal Offset	1.55	0.72
15 Degree Angle	-	0.76

SOFTWARE DESIGN

BrachI/Oplexus, an open-source graphical user interface (GUI) designed for myoelectric prosthesis control [11], enables the EMG signal interpretation and end effector motion. A microcontroller (Arduino Uno, R3) controls the mechanotactile tactors and grip force sensors. Data logging capability is enabled at a frequency of 50 Hz. A custom GUI (Visual Studio, 2015) was created to communicate with the microcontroller for quick customization of tactor parameters.

CONCLUSIONS AND FUTURE WORK

A lightweight, modular simulated prosthesis was developed with integrated modality and somatotopically matched mechanotactile feedback. Grip force sensors were compared before and after being encapsulated in a compliant material under various loading conditions. In all non-standard loading conditions, the encapsulated sensors outperformed the baseline sensor. This device will help enable researchers to study feedback and control techniques in myoelectric prosthetics by providing a reliable test apparatus that easily allows for the manipulation of various parameters.

Future work includes evaluating the performance of the MSP to ensure that the device is an accurate representation of a myoelectric user and evaluate the effectiveness of various sensory feedback techniques. More modular components, such as alternative feedback devices of various modalities, could be designed to fit onto the device. The device is currently tethered to a one-meter long power cable, which may be restrictive for some studies. A wireless version of the MSP would make the device more flexible.

ACKNOWLEDGEMENTS

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DEVELOPMENT OF A UNIVERSAL TRANSRADIAL FITTING FRAME

Diana Pressney, CPO[1] & Laura A. Miller, CP, Ph.D.[1,2]

[1] Shirley Ryan AbilityLab, Chicago, [2] Northwestern University

ABSTRACT

The Center for Bionic Medicine conducts research on upper limb prosthetic devices, components, and technologies. In order to maximize the amount of human subject research able to be completed, a universal fitting frame for subjects with transradial amputations is needed. This paper will outline the development of a universal transradial fitting frame which has been in use for approximately one year at the Center for Bionic Medicine, Shirley Ryan AbilityLab, Chicago. This device is easy to assemble, easy to use, and allows for testing a wide variety of prosthetic wrists and hands.

INTRODUCTION

There are approximately 100,000 people living with major upper limb loss in the United States [1]. Loss of a hand has been shown to cause great functional loss [1]. A prosthesis can improve functional use as well as quality of life for people living with limb loss [2]. Therefore, there is a strong need for research into prosthetic components and their effectiveness.

At the Center for Bionic Medicine (Shirley Ryan AbilityLab, Chicago, IL), research into device development, outcome measures, and new technologies are being completed on a yearly basis. In order to improve feasibility of human subject research, having a device to easily test componentry on any transradial limb is necessary. The current process for in-lab testing is to create a custom plastic socket and mock up a temporary prosthesis. This process requires at least two visits for each subject to ensure proper fit of socket. If this timeline could be reduced, study times could be reduced, and devices evaluated at a faster rate ultimately improving clinical knowledge.

The purpose of this project was to develop a universal fitting frame that would fit the majority of transradial residual limbs in order to use any available prosthetic wrist/hand combination for research purposes. The project had five main product scopes that needed to be addressed: 1. It needed to be adjustable to fit most transradial limb lengths, 2. It needed to be adjustable to fit any girth transradial limb, 3. It needed to be adaptable to a liner integrated with electrode domes, 4. It needed to be adaptable to work with the different commercial and research myoelectric wrists/hands currently used in experiments, 5. It needed to be able to withstand forces necessary to complete standard outcome measures including box and blocks, clothespin test, SHAP, and ACMC.

DEVELOPMENT OF THE FITTING FRAME

<u>First Prototype</u>

Based on the above project goals, we proceeded with development of the fitting frame in three main components. The first was the structure itself; the structure had to be light weight and adjustable. A channelled aluminium bar (6063 Aluminium Rectangular Tube, $^{1}/_{16}$" wall thickness, ½" high, 1" width) was used in conjunction with a piece of aluminium bar stock (6061 Aluminium, ¼" thick, ¾" wide). A series of holes ($^{10}/_{32}$") were tapped into the channelled bar and two set screws were then used to hold the solid aluminium bar in place inside the channel. This channel design allowed for adjustment in length for different length residual limbs. Two channelled bars and two solid bars in different lengths (short and long) such that they could be interchanged depending on length needs.

The second component was the suspension. A flexible cuff made from a combination of flexible and rigid lamination (Paceline Nano Matrix Resin) was used to grip the muscle belly just distal to the elbow joint. Two cuffs were made in two sizes to accommodate for small and large limbs. The cuffs purpose was to provide minimal support for the frame. The main source of suspension came from an adjustable ratcheting mechanism (Revoflex BOA kit). A BOA ratcheting dial, commonly used in prosthetic and orthotic applications, was embedded into two pieces of laminate lined with plastazote to create a sort of clamp system. This allowed for adjustability depending on the limb circumference and provided excellent suspension between the patient's limb and the posterior bar. For suspension between the liner and the wrist unit, a magnet connection (High-Pull Rare Earth Magnetic Disc with countersunk mounting hole, $^{7}/_{8}$" diameter) was utilized. One magnet was inlayed to the wrist connector and one was attached to the bottom of a locking liner. Finally, anterior straps were added to act as a counterforce system to offset weight of whichever hand/wrist combination was connected. These components can be seen in Figure 1.

The last major phase of development was the wrist connection. Custom 3D printed pieces were created in Solidworks and printed with a uPrint SEplus printer using ABS. The first piece created was a universal connection that would connect the structure of the fitting frame to whichever wrist unit was to be utilized. This piece included an extrusion which would fit into the channelled aluminium bar, a space for the magnet, and three wings which would be used to connect to all distal componentry. The second piece was developed to house the specific prosthetic wrist unit. This second piece could be easily altered to different wrist units by changing dimensions. The only thing that needed to stay the same was the three wings which are used to connect to the top connector. See Figure 2 below for a sketch of the completed pieces in Solidworks showing how the two parts mate together. Spacers of various heights with the same connection pattern were also fabricated so that the overall length could be extended to match the contralateral limb when using a shorter wrist system.

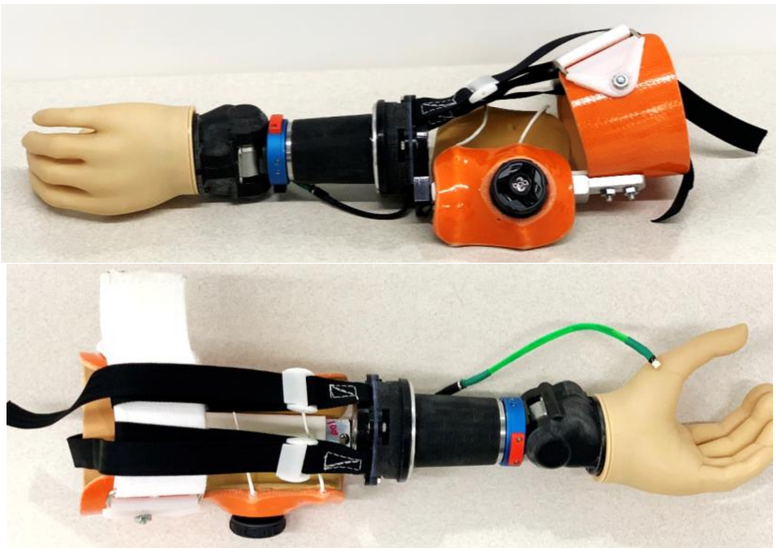

Figure 1: Photograph of assembled fitting frame (first iteration) with BOA clamp design.

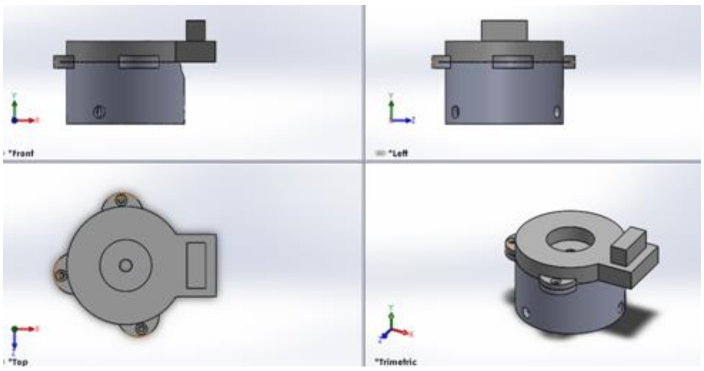

Figure 2: Solidworks image showing both custom 3D printed pieces as they would work together to connect the frame to the prosthetic hand.

Second Prototype

After trial use of the initial prototype, several modifications were made. Originally, the counterforce straps were attached using Velcro that attached to the cuff strap. However, to improve line of pull, the design was altered to a buckle attachment. The second change was the switch from the custom BOA clamp system to an off the shelf BOA strap mechanism. This makes it easier to duplicate the device as well as reduces overall bulk.

Figure 3: Photograph showing all components of the second iteration of the universal transradial fitting frame including the new counterforce straps and BOA strap. In the center is the assembled device with extra components surrounding.

Figure 4: Photograph of all components of the second iteration of the universal transradial fitting frame broken down to show each part individually.

STRENGTHS AND LIMITATIONS

The biggest strength of this fitting frame is its adaptability. It is possible to quickly configure the device to fit different lengths and shape residual limbs. This allows for ease of set-up for experiments and reduces time to actual testing. The device is lightweight and simple in design. It is easily replicated from materials that are fairly easy to acquire. The most unique component is the 3D printed wrist connection which does require access to a 3D printer.

While the fitting frame does work well for most situations, there are a few limitations to this design. The major limitation is that it does not work with very short residual limbs (length under 10cm from lateral epicondyle to distal end of residuum). The second limitation is that it is not ideal for heavy components, especially when on a short residual limb. It relies on dacron straps to act as the counterforce for the wrist and hand componentry. If the patient presents with a short residual limb, the lever arm is shortened and more force is needed from the counterforce straps to maintain alignment.

CLINICAL AND RESEARCH IMPLICATIONS

The universal transradial fitting frame is easy to use, adaptable to fit any residual limb length and shape, and has been used to test a variety of current prosthetic wrists and hands. To date, the fitting frame has been used on seven subjects with success. Four different wrists have been used; two commercial wrists (Ottobock and MotionControl) and two research wrist systems. EMG has been connected to the system via Motion Control snap electrodes in the locking liner and reliable and consistent signals were obtained. A sample image of a subject with a transradial amputation wearing the fitting frame is shown in Figure 5. The design is simple and easy to replicate in any research or clinical application.

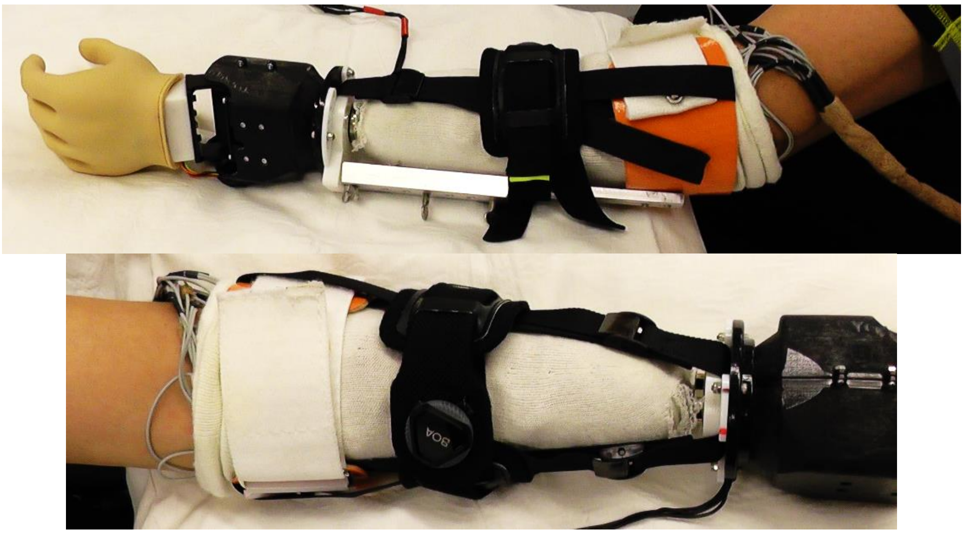

Figure 5: Photograph of a transradial subject wearing the fitting frame.

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Externally Powered Prosthetic Wrist Flexion Device

Edwin K. Iversen [1], Jeff Christenson [1], Gregory Jacobs [1], Scott Hosie [1]
[1] Fillauer LLC

BACKGROUND Externally powered prosthetic wrist rotators have been used by individuals with upper limb loss for over 20 years [1, 2]. The recent availability of an externally powered prosthetic wrist flexion device raises these two questions: 1) what are the functional benefits for individuals with upper-limb deficiencies of an externally powered wrist flexion device, and 2) are there advantages to externally powered wrist flexion over externally powered wrist rotation?

AIM
It is the aim of this paper to help physicians and prosthetists to determine when an externally powered wrist flexion device might benefit their patients with upper-limb deficiencies.

METHOD
A wrist flexion device for a remnant limb that has an intact elbow and shoulder was designed and evaluated. Three steps in the process were: 1) kinematic analysis comparing the functionality of powered flexion versus powered rotation 2) development of design objectives, and 3) obtaining of user feedback from individuals using powered flexion prostheses.

A kinematic analysis showed the hand orientations that are possible with different types of wrist. For instance, the analysis showed how well a user could use such objects as a flashlight, a fork, or a personal cleaning device. We compared the use and viability of a Powered Flexion Wrist system, a system with no wrist, a system with a wrist rotator, and a system with both wrist rotator and wrist flexion. The analysis was performed using standard joint space calculations.

Following the kinematic analysis, design objectives for the powered flexion unit were developed based on input from prosthesis users, industry experts, and the literature. These design objectives drove the development of the Powered Flexion Wrist (PFW) unit.

Once designed and developed, two rounds of field trial participants were recruited for PFW evaluation, for a total of eight responses. Participants' evaluation of the PFW was performed through a questionnaire asking the participants to evaluate different aspects of the device on a scale from -2 to 2, which -2 being highly unsatisfied and 2 being very satisfied.

RESULTS

Kinematic analysis
The kinematic analysis of the four wrist systems shows that a flexion wrist enables a different type of workspace from that of a wrist rotator [3]. The analysis also shows that a system with both wrist rotation and wrist flexion enables the most complete workspace (Figure 1). If the user and prosthetist need to choose between either flexion or rotation, they should choose the wrist which aligns most with the individual's desired functional outcomes. However, a wrist with both flexion and rotation will enable more achievable orientations of the hand, and therefore significantly greater and more natural functions.

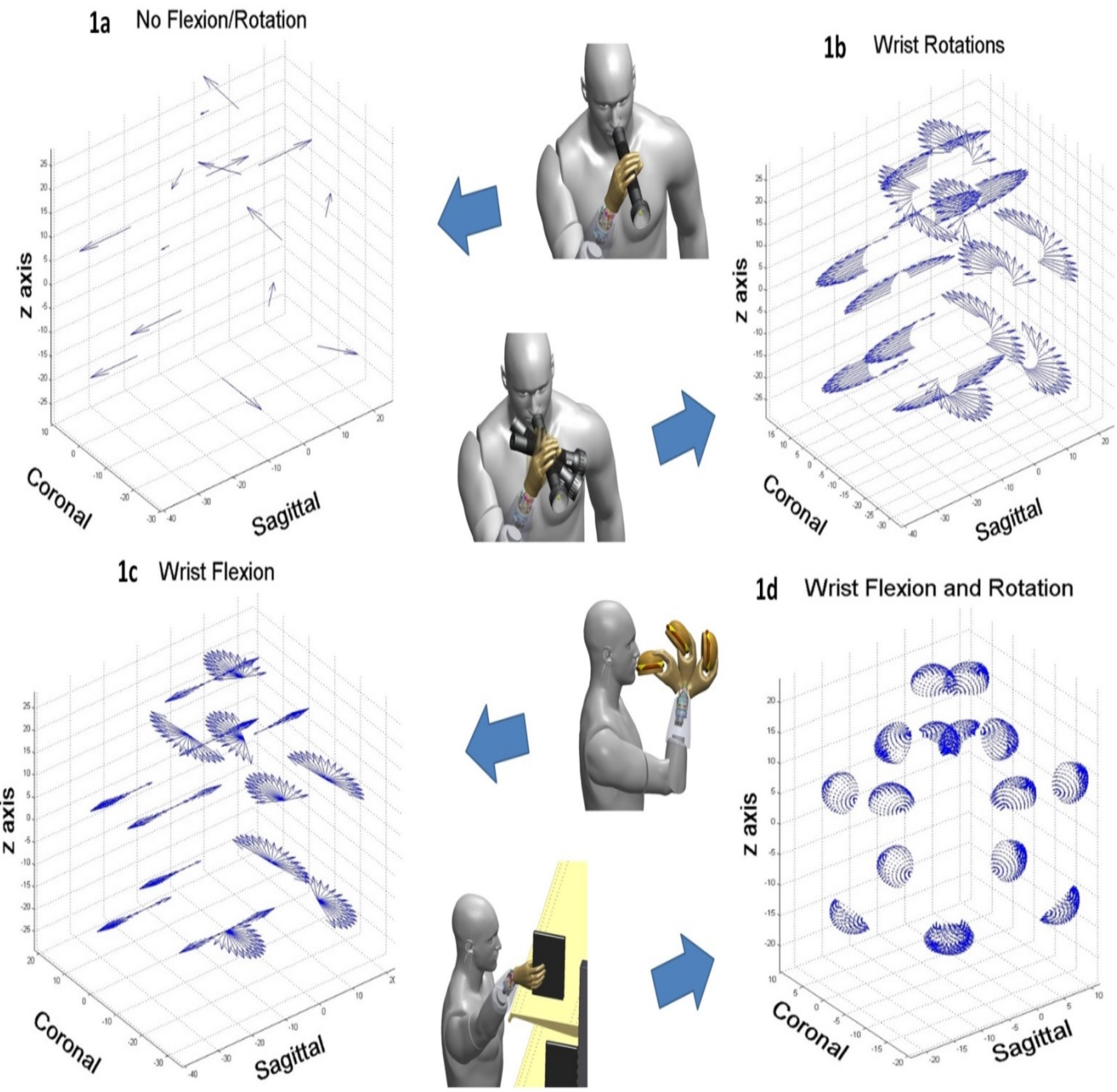

Figure 1: Results of kinematic analysis of 4 different wrist systems. 1a shows the workspace of a system with no wrist. 1b shows the workspace of a system with a wrist rotator. 1c shows the workspace of a system with a wrist flexion device. 1d shows the workspace of a system with both wrist rotation and wrist flexion. Note that the vectors or arrows in 1a, 1b, and 1c show the possible orientation of the axis of a cylindrical object, such as a flashlight, grasped transverse in the hand. However, in figure 1d, which shows that the user can orient a grasped object in virtually any direction, only the end points of the possible vectors are shown for clarity.

Design objectives

The design objectives and achieved results of the PFW are shown in Table 1, along with the specifications of the powered flexion for comparison. Notice that the weight achieved is lower than the target, but still higher than the wrist rotator alone.

	Target	Achieved	PFW+ Wrist Rotator
Length (not including QD) (mm)	57	66	70
Diameter (mm)	48	48	47
Weight (grams)	340	259	259 + 143 = 402
Active Torque Max (Nm)	2.8	2.3	1.7
Passive Torque Max (Nm)	2	2.3 (1.7)	Na
Rotational Travel (Degrees)	145	153	360
Rotational Speed (rev/s)	0.5	0.5	0.533
Bluetooth	yes	yes	yes
IPX7 tested	pass	pass	pass

Table 1: Design objectives and results of the Powered Flexion Wrist. As a reference, the specifications of the PFW + Wrist Rotator, also produced by Motion Control, are also shown.

Field Trial
A total of eight individuals, in two cohorts, evaluated the PFW for at least two months. After the first trial, the wrist was refined before being tested by the second cohort. At the end of the evaluation period, each individual provided feedback through a survey. Figure 2 shows the results of the survey [4,5]. The data shown is the sum of all user responses, separated by cohorts.

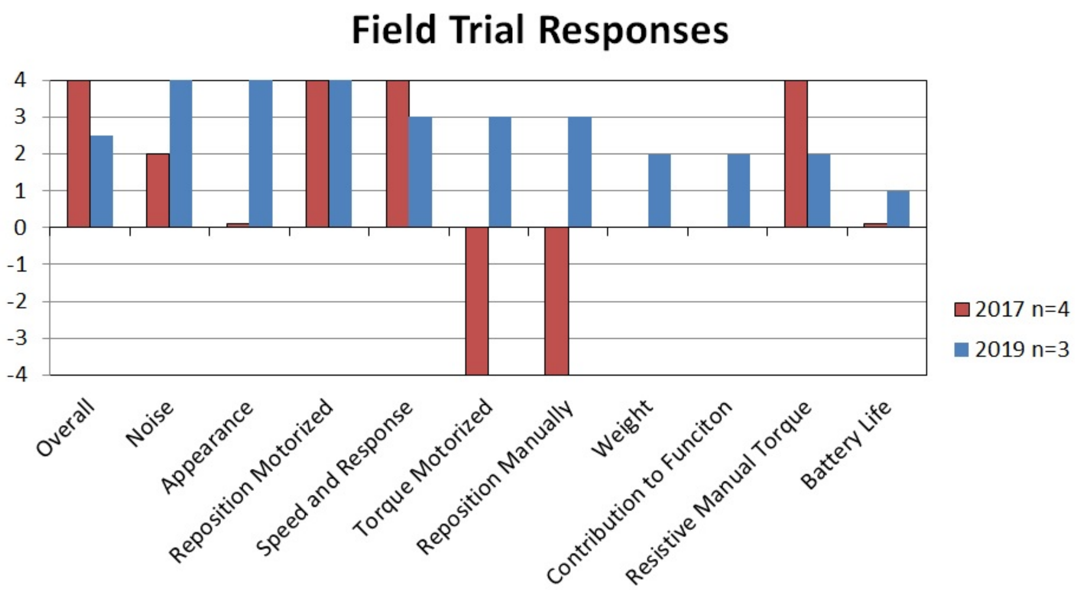

Figure 2: Field Trial responses of the two cohorts.

The first cohort was dissatisfied with the motorized torque and the manual repositionability of the PFW. Between the two field trials, the motor torque and the manual repositionability were improved, as can be seen by the second cohort evaluation data. Overall, prosthesis users evaluated the PFW favourably.

In the process of administering the final prosthesis user survey, other factors were discussed by individuals which are important to note. Two of the final prosthesis users desired the PFW to be shorter. Two of three individuals also mentioned a decrease in system battery life when using the PFW, especially with a pattern recognition system.

DISCUSSION

From the kinematic analysis, it is clear that one type of wrist does not fully define the workspace. The data shows that the PFW is beneficial for tasks requiring access to the midline of the body, such as eating and dressing, or picking things off the floor or table (Figures 3). Other tasks, such as unscrewing a bottle or turning a key, are more easily accomplished using a wrist rotator. An ideal solution would be to have both a rotator and a PFW, since the workspace is greater. However, length, weight, battery life, and the need to control so many degrees of freedom must be taken into consideration for different individuals.

Figure 3: Individual executing tasks using the Powered Flexion Wrist.

If only one wrist function can be integrated into a prosthetic system, the prosthetist should recommend the wrist which best matches the desired functional outcomes of the individual.

Although the weight of the PFW is heavier than the wrist rotator alone, it is interesting that the field trial participants did not rate the wrist as being too heavy. One respondent made the point that if the device is functional, the weight is secondary.

The first set of feedback motivated a design change in the transmission, which increased both the passive positionability and the active torque of the system. The second set of feedback indicates that the current design is acceptable.

CONCLUSION

The kinematic analysis and the results from user questionnaires clearly show that a Powered Flexion Wrist offers potentially significant functional benefits for individuals with upper-limb loss. When choosing an appropriate wrist, one must consider the types of tasks desired to be performed and the person's functional workspace.

DISCLOSURE

Fillauer manufactures externally powered and passive flexion and rotation devices.

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Improved Prosthetic Functionality Through Advanced Hydraulic Design

Bjørn Olav Bakka[1], MSc and MBA, Norway, bb@hy5.no
Christian Fredrik Stray[1], BSc and MBA, cf@hy5.no
Jos Poirters[1], jp@hy5.no
Ole Olsen[1], MSc, ole@hy5.no
[1]Hy5 Pro AS, Bygning 100, Raufoss Industripark, 2830 Raufoss, Norway

ABSTRACT

Hy5 met its research objective of designing a hand prosthesis to fill the gap between standard myoelectric grippers and premium, bionic-like hand prostheses. Our approach applied state-of-the-art hydraulic actuator technology with functionality embedded in advanced 3D printing of titanium and plastics. As a result, the opening and closing of the hand is myo-electrically controlled and compatible with industry standards while the hydraulics enable an adaptive and independent pressure build-up on the fingers as they grasp an object. This design mimicking realistic hand gripping without requiring one motor per finger as in bionic-like prosthesis.

Testing concluded that the MyHand prosthetic hand manages all grips (pinch, power, fist, tripod and point) as intended and works as a substitute for a missing hand. Users also responded very favourably to the innovative emergency release button, an added safety feature. The users were attracted by the simplicity and sturdiness of Hy5, which promises a reliable product with low life-cycle cost.

INTRODUCTION

Advancements in technology can result in overly complex designs leading to underutilized features. Advanced prosthetic devices are no different, specifically with overly complex hand prostheses where highly technical designs may lead to increased weight and cost, along with reduced reliability and usability. Additionally, when a person experiences an amputation, they face staggering emotional, practical, and financial lifestyle changes.[1] Following such an event, the person typically requires a lifetime of costly prosthetic device(s) and services, reduced physical activity, and difficulty with community reintegration and full participation in social life. Losing a limb has been found to dramatically change a person's sense of body image and consequently self-image, which has, in turn, been associated with a person's satisfaction with life.[2] An upper-extremity (UE) prosthesis is considered among the most challenging prosthetics devices to use, both from a functional and a control perspective.

Compared to the typical UE prosthesis, the biological human hand is complex device. With 38 muscles[3], 27 bones[4], 21 Degrees of Freedom (DOFs), thousands of touch sensors, a human hand is direct skeletally-attached and weight-bearing, capable of swift movements, and designed for life. Alternatively, a typical prosthetic hand has few DOFs, no sensors, its distal weight is supported only through a socket, it is much slower and imprecise than a biological human hand and is regularly in need of service and repair. The biological human hand is controlled naturally through afferent sensory input and efferent motor output signals of the Somatic Nervous System, while a myoelectric prosthetic hand is controlled through learned intentional, yet often unintuitive muscle contraction.

The human hand is used as an indispensable tool in daily life. There are several reasons why the human hand should not or cannot be copied in order to produce effective end effectors and terminal devices[2] as current state of the art in engineered systems cannot achieve a comparable level of complexity and performance in the same size package. Due to the many reasons the full spectrum of human hand capabilities cannot be practically achieved in a prosthetic hand, some smaller subset of those must be chosen. Several studies have concluded that a small number of grasp types comprise the majority of those used.[2] Other studies have shown that weight, cost and reliability is a concern with higher preference by users over independently moving fingers.[5]

First demonstrated in the 1940s, myoelectric prosthetic hands rely on electrodes applied to the skin to detect and translate muscle pulses drive a device actuator. The actuator can be hydraulic (i.e. pump and cylinders), electromechanical (i.e. motor and gears) or pneumatic (i.e. compressed gas). The DOF is usually limited to only one – open or close hand. The 1940s myoelectric control technology is still the most widely-used control method, while technology achievements have made the components lighter, cheaper and more reliable.

Several anthropomorphic multiarticulate prosthetic hands have been developed and introduced onto the international market in the previous two decades.[6] Common to all of them is a complex design with high number of DOF and actuators that still rely on two-sensor myoelectric control with the basic "open" and "close" commands. This is a clear example of overdesign by applying new technology indiscriminately, and often requiring the user to switch between hand operation modes by means of buttons, apps or muscle co-contractions. This complicates operation by increasing cognitive stress and training needs and results in underutilization of the capabilities of the prosthetic hand. Additionally, most of these advanced prosthetic hands still bear the cost of increased weight, reduced reliability, and reduced affordability.

The situation is that the current prosthetic technology provides limited options for amputees: patients are provided with either standard utilitarian myoelectric grippers with limited functionality, or advanced and expensive bionic-like hand prostheses. Each of these choices results in underutilization or inadequate functionality, or both.

Hy5 has integrated a simple design with lightweight materials and advanced motion control and flexibility, resulting in a prosthetic hand that improves utilization and functionality for daily life. The MyHand design addresses the critical functional and economic gap that exists between body-powered and relatively simple myoelectric devices, and high-cost anthropomorphic multiarticulate prosthetic hands.

METHODS

The technology that led to Hy5's "Improved Prosthetic Functionality Through Advanced Hydraulic Design" was initiated more than 15 years ago. Our development path began with the idea to replace error-prone electric motor actuators with hydraulic actuators in dolls in an amusement park. Since then, the work has evolved, inspiring us to make a better life for people living with upper extremity amputation, with the vision of "Giving the World a Helping Hand".

Figure 1: Hy5's vision of "Giving the World a Helping Hand" - MyHand testing with newly amputee.

Hy5's design employs a hydraulic actuator, which is one of several possible actuators, or "muscles", that can be used in prosthetic limbs. In 1985 it was stated that "electro-hydraulic systems may be used in the future because they have the potential advantage of developing high torque in small actuators" [7]. Hydraulics is well-proven technology with documented benefits for prosthetic lower-limbs. Several research projects have resulted in hydraulic actuated prosthetic hand prototypes. Examples include the "Fluidhand" developed in Karlsruhe, Germany and a mesofluidic hand developed in Oak Ridge, USA.[8] However, Hy5 is the first company that is designing, producing, and selling a hydraulic prosthetic hand.

The MyHand prosthetic hand integrates several innovative features, some of which are patented. The palm unit integrates the electrical motor, hydraulic pump,[9] cylinders and piping and is 3D printed for low weight, low cost, high flexibility and high complexity. The hydraulic pump is a single high-volume and high-pressure integrated pump.[10] The high-volume pump provides the high non-resistance opening and closing speed, while the high-pressure pump provides the high gripping force. The digits are closed by wires being actuated by the palm cylinders. The digit mechanism is a force balancing mechanism[11] enabling the digits to close on objects regardless of their shape. Major parts of the digits are 3D printed in titanium for low weight and high durability.

Figure 2: MyHand Advanced Hydraulic Design

The opening and closing functions of the MyHand prosthesis are myoelectrically controlled. The prosthesis uses a single motor to control three hydraulic cylinders. Each hydraulic cylinder controls the digits of the thumb, index, and middle finger by means of the mechanical wire solution in their respective knuckle joints. This enables an adaptive and independent pressure build-up on the thumb, index and middle fingers while the ring and pinkie fingers move together with middle finger as they grasp an object, thus mimicking realistic hand gripping without requiring one motor per finger.

Figure 3: MyHand Production Testing

The MyHand device design specifications demonstrate its impressive performance: a maximum power grip of 120N, maximum tripod grip of 60N, maximum static load of 40kg, the maximum time to close is 1.2 seconds and weight is 580g.

Table 1: MyHand Specifications

Performance specification	
Maximum power grip	120 N
Maximum tripod grip	60 N
Minimum time to open/close: power grip	1,5 Sec
Minimum time to open/close: tripod grip	1,5 Sec
Maximum static load: hook grip	40 Kg
Maximum load individual finger – hook grip	20 Kg
Fingertip extension load	8 Kg
Weight	580 g
Size	7 ¾

RESULTS

The Southampton Hand Assessment Procedure (SHAP)[12] is designed to measure a hand's functional range. The procedure was developed in 2002 at the University of Southampton to assess the effectiveness of upper limb prostheses. The SHAP test consists of a series of manipulations of both lightweight and heavyweight abstract objects intended to directly reflect specific grip patterns while also assessing the strength and compliance of the grip, followed by 14 Average Daily Life (ADL) tasks.

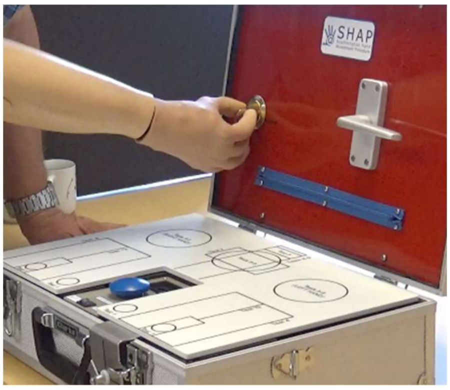

Figure 4: SHAP briefcase with gloved MyHand

In late 2017 the SHAP was conducted internally with the MyHand prosthetic hand used by 21 subjects comprising 20 males, and 1 female users, ages 27 to 65[13,14]. The testing revealed some variations in how the users managed to control the device. Users with limb-difference from birth generally have longer experience with handling a prosthesis and managed to control the MyHand hand quicker than users amputated later in life. Some managed to control MyHand instantly, others needed more time to get accustomed to it. An orthopedic specialist observed the testing together with Hy5 employees. Both the orthopedic specialist and the user were interviewed after testing.

Figure 5: MyHand SHAP testing

The SHAP testing concluded that the MyHand device manages all grips (pinch, power, fist, tripod and point) as intended and works as a substitute for a missing hand. General user feedback and specific results from the SHAP testing have been positive.

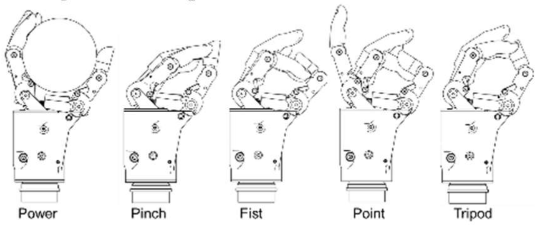

Figure 6: MyHand Grip Patterns

The users expressed specific satisfaction about the ability of the MyHand to adopt to and grip complex objects. All users were very positive to the extra safety, accomplished with the emergency release button on the Hy5. The emergency release button releases all hydraulic pressure on the fingers, which will then open by themselves or may easily be forced open. The emergency button may prevent the hand from breaking when the locked around an object, or the battery is empty, and the hand is forced open by breaking it. None of the users have seen this feature on any other hand prosthesis today. The users were attracted by the simplicity and sturdiness of MyHand promising a reliable product.

DISCUSSION

Analysis of the SHAP testing shows that the grip patterns of the MyHand prosthesis allow recovery of up to 30% of total gripping functionality required for activities of daily life (ADL's) compared to standard grippers. This is an important part of the MyHand value proposition.

Figure 7: MyHand Power Grip and Fist Grip

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Further analysis showed that user functionality achieved with the MyHand prosthesis is comparable to that of most advanced bionic-like prosthesis users. Functionality in the advanced bionic-like hand requires significant training, cognitive attention and risk of faulty functionality. Access to MyHand gripping patterns is intuitive with less training and cognitive attention.

One benefit of the MyHand is the simplicity and sturdiness of the hand which supports a reliable product resulting in low life-cycle costs. For users this translates into less time lost to breakage or servicing, minimizing time spent without the use of the hand. Being rugged, the MyHand hand can be employed in activities and environments where other hands will break, improving quality of life by enabling new lifestyles. Whether the user pays for the device personally, or with the use of insurance, low life-cycle cost simply means fewer budget restraints and the ability to service more people.

The MyHand prosthetic hand has received regulatory approval in Europe, US, Australia and Canada.

CONCLUSIONS

Hy5 has designed a prosthesis to fill the gap between standard myoelectric grippers, and premium, bionic-like hand prostheses. This technology offers cost-effective advanced motion control and flexibility with critical functionality. Hy5 will break critical barriers for user comfort, directly addressing the existing needs for lighter and faster hand prostheses. Providing the general public with a wider variety of options allows individuals the best fit to their lifestyle, and an improving quality of life.

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Perspectives on Leveraging Advancements in Adult Upper Limb Prostheses to Improve Pediatric Device Acceptance

Marcus A Battraw[1], Wilsaan M Joiner[2], Michelle A James, MD[3], Jonathon S Schofield[1]

[1]Department of Mechanical and Aerospace Engineering, University of California, Davis, Davis CA, USA
[2]Departments of Neurobiology, Physiology and Behavior; Neurology, University of California, Davis, Davis CA, USA
[3]Shriners Hospital for Children - Northern California, Sacramento CA, USA

ABSTRACT

There are many complex factors that contribute to whether a child with a congenital limb difference will use or abandon their prosthetic limb. When compared to adults with traumatic amputations, children with limb deficiencies are less likely to use a prosthesis, and many of their challenges are unique to being a child. Ultimately, for a child to adopt their device, it must facilitate the effective performance of daily activities and allow the child to be treated the same as their peers. Although numerous pediatric devices are available, they often fall short of these criteria by offering a single open-close grasp and/or non-anthropomorphic appearances. However, when looking to the field of adult prosthetics, multi-articulating myoelectric hands can provide multiple grasping configurations and have the benefit of a more 'hand-like' appearance. If these designs are adapted for pediatric users, their advantages have the potential to improve device acceptance. In this paper we provide a critical assessment of the state of upper limb prostheses for pediatric populations. Furthermore, we suggest ways that we may leverage recent advances in adult myoelectric devices to begin removing the barriers to pediatric device adoption. Finally, we discuss how current challenges in the adult myoelectric field must be considered to effectively translate this technology.

INTRODUCTION

It has been estimated that congenital transverse below elbow deficiencies occur in approximately 1 of every 10,000 live births [1]. For these children, a passive prosthesis may be prescribed as young as 6 months of age and active devices as early as 18 months [2]. The use (and/or abandonment) of these prescribed devices is a multi-dimensional challenge. Parents play a vital role in the decision-making processes that influence use and adoption while their child is too young to make these decisions for themselves. It is common for guardians to view their child's limb difference as a deficiency that needs to be addressed with an artificial limb [3]. However, when the child comes of age to make their own decisions, prosthetic abandonment quickly become more common [4].

Much like adult upper limb (UL) prosthetic users, device abandonment is a common occurrence; however, in pediatric populations it is a more prevalent and pervasive issue [5,6]. In 2007, Biddis and Chau reviewed 25 years of literature and suggested that adult prosthetic abandonment rates varied from 26% for body-powered devices to 23% for electric [5]. They further suggested that children face far more complexity in the prosthetic arena, resulting in abandonment rates for body-powered and electric prothesis at 45% and 35%, respectively [5]. Regardless of age, the key factors that ultimately impact use and acceptance of prostheses can be placed into three categories: social, prosthetic/technical, and clinical/personal factors [5]. Appearance, functionality, and weight can be further isolated as being particularly relevant to children [4,7], and *prosthesis usage is ultimately contingent on providing sufficient functionality and cosmesis to allow the child to be treated the same as their peers (social integration)* [8].

In this paper, we critically assess the state of UL prostheses for pediatric populations with congenital limb differences. Furthermore, we summarize the prevailing technical and social challenges that prevent the wider spread adoption of these devices. Finally, we suggest ways that we may leverage recent advances in adult myoelectric prostheses to begin removing the barriers to prosthetic acceptance and reduced abandonment rates in pediatric populations.

CURRENT PEDIATRIC PROSTHESES OPTIONS

Current pediatric prosthetic devices will either be passive (cosmetic), body powered, or myoelectric devices. Although passive devices may often appear more life-like or anthropomorphic in appearance, they lack critical functionality as they do not provide the ability to actively grasp. Body powered prostheses may offer many attractive qualities including minimal weight, cost, ease of control, and robustness. However, most of these devices are limited to a simple open-close grasp which inherently requires the user to employ compensatory strategies to achieve many daily grasping tasks. When coupled with their often-non-anthropomorphic appearances, body powered devices simply do not meet the functional and cosmetic demands to promote social integration. Current pediatric myoelectric devices typically offer a single degree of freedom (open-close) terminal device and in some cases wrist rotation. They provide the benefit of control using the muscles native to the affected limb which may remove the need for additional cables and harnessing as well as body/shoulder movements to control the terminal device. However, myoelectric devices come with a number of practical challenges including increased weight, often reduced robustness [7], slower actuation of the grasper, and challenges achieving consistent control.

Presently, body-powered prostheses are often preferred to myoelectric devices when performing functional tasks [9]. Crandall et al. surveyed the satisfaction of pediatric patients and their parents in relation to using their prosthetic device during daily activities. In their cohort of 34 users between the ages of 1 to 12 ½ years, body-powered devices were able to achieve more functional tasks to the users' satisfaction when compared to passive and electric devices. Surprisingly, in a long-term follow up more than a decade later, most of these same patients were using a passive device [9], suggesting that *the current single degree of freedom grasping function provided by an active prosthesis offers limited benefit relative to no-grasping function at all*. As a result, these patients opted to use a passive device that, although less functional, may provide improved cosmesis to help facilitate social integration. Further empathizing the magnitude of these challenges, in a survey-based study of 489 children with a unilateral congenital below-the-elbow deficiency (321 prosthesis users and 168 non-users), James et al. found no clinically relevant differences between prosthesis users and non-users in validated measures of functional outcomes and quality of life [10]. Furthermore, when investigating the performance of various daily tasks, they found non-users scored themselves higher than prosthetic users. This guided their conclusion that pediatric prostheses may provide cosmetic benefit for social acceptance or may be useful tools for specialized activities, but at present, they do not appear to improve patient function or quality of life [10].

GRASPING PATTERNS AND DAILY FUNCTION

Unlike the single degree of freedom grasping function offered by current active pediatric prostheses, healthy intact hands are incredibly dexterous with 27 degrees of freedom [11]. Although it is possible to achieve a multitude of complex postures with this available dexterity, most activities of daily living are performed using a limited number of common hand grasp configurations [12,13]. In fact, it has been suggested that nearly 80% of common daily tasks can be accomplished with as few as 6-9 standard grasp configurations [12]. Therefore, we suggest that a significant functional benefit may be provided to pediatric prosthetic users if their devices offer multiple grasping configurations to more effectively accommodate the performance of daily activities. This challenge is not unique to pediatric prosthetic users and closely parallels a very active body of work being performed with adult amputee populations.

LOOKING TO ADULTS

In recent years, multi-articulating adult myoelectric prosthetic hands have become increasingly available. There are now numerous commercially available options with individually actuating digits that can achieve a multitude of common grasping configurations [14]. Table 1 adapts data from a metanalysis of hand grasp literature [12]. Here, we list the top 6 most frequently used grasp configurations by intact hands in daily activities and compare them to the capabilities listed in manufacturers' literature of prevalent adult multi-articulating myoelectric hands [15–20]. Nearly all the top 6 hand grasp patterns are capable of being achieved with these current adult devices. Beyond their added function, an additional advantage inherent to their hand-like designs is that these prosthetic devices also appear more anthropomorphic or life-like than many of their body-powered hook-and-cable counterparts.

Together the added function of multiple grasp patterns and the improved cosmesis of adult myoelectric hands has the potential to address two crucial factors that influence pediatric prosthetic use. In fact, multi-articulating prosthetic hands are beginning to emerge in the pediatric field. For example, the Vincent Young 3 (Vincent Systems, Karlsruhe, Germany) is sized for children age 8 and up, is capable of 13 individual grasp patterns, has four wrist options, and is made of lightweight materials. However, these devices have only started to become available and have yet to see widespread adoption. There are a number of practical and clinical challenges that will likely first need to be addressed.

MOVING FORWARD WITH PEDIATRIC PROSTHESES

There are many considerations and barriers to multi-articulating myoelectric devices that may be both common and unique to pediatric and adult populations. Device cost is a significant and prohibitive barrier for both populations. However, it is a distinct obstacle for pediatric patients as their limbs and body are ever-growing. Therefore, unlike adults where purchasing a single terminal device may be a long-term investment, the cost of children's devices must reflect the fact that a child will likely outgrow a device in a few short years and multiple devices will be purchased over their childhood.

Table 1: Commercially Available Adult Multi-Articulating Adult Prostheses and Most Frequent Grasp Configurations used in Daily Activities.

Prostheses \ Grasps	Power Grip	Precision Pinch	Key Grip	Tripod	Precision Disk	Prismatic 2 Finger
BeBionic	✓	✓	✓	✓	✗	✓
i-Limb*	✓	✓	✓	✓	✗	✓
Michalangelo Hand	✓	✗	✓	✓	✗	✓
Vincent Evolution 3	✓	✓	✓	✓	✗	✓
Luke Arm	✓	✓	✓	✓	✗	✓
Taska Hand*	✓	✓	✓	✓	✗	✓

Note: Prosthesis grasp data derived from available manufacturers' literature Activities [15–20], and Grasp configurations adapted from Feix et al. [12].
**prostheses allow for custom grasps to be programmed.*

Furthermore, the growth of a child also poses a unique barrier to achieving consistent myoelectric control. As affected limb proportions changes so will socket fit and the contact of electrodes over muscle control sites. This may result in diminished, inconsistent, or intermittent device control. In addition, most pediatric patients are born with their limb difference. Effective contraction of muscles on the affected side will inevitably require structured training and learning prior to being used for prosthesis control. However, here again we may look to advancement in the adult prosthetic field to mitigate some of these barriers. Commercially available control systems that employ myoelectric pattern recognition may be a viable option in alleviating some of these control challenges and facilitating intuitive control over multiple grasp configurations. Similarly, emerging experimental techniques that leverage ultrasound-based control or force-myography may also provide avenues for further investigation [21].

Finally, robustness and 'bulk' of a myoelectric hand have unique and interconnected implications to pediatric prosthesis use. When comparing activities of daily living between adults and children, we suggest that children will likely require a more robust device to facilitate the physical nature of childhood play. Robustness typically comes at the cost of a more rugged design with increased weight. Children are more affected by the weight of the device [22] as they are smaller and do not possess the same strength as a grown adult. Furthermore, multi-articulating prosthetic hands are innately heavier as they require motors and additional mechatronics to actuate digits. This additional componentry must also be housed within the device which may impact its overall size. Therefore, as multi-articulating pediatric prosthetic hands continue to emerge, significant attention must be dedicated to developing devices that incorporate lightweight materials and creative 'low-bulk' design principles.

CONCLUSIONS

The factors that contribute to the use and acceptance of pediatric UL prostheses are complex and abandonment is highly prevalent. There have been many advancements in adult UL prostheses that have yet to be leveraged which may positively impact the pediatric arena. By adapting the capabilities of adult multi-articulating myoelectric prostheses, we can begin addressing some of the crucial factors that are contributing to the disuse of pediatric devices. However, there are numerous challenges that are unique to this patient population that must be carefully considered to inform and shape the development of future multi-articulating pediatric prosthetic limbs.

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REALISTIC DESIGN CONSIDERATIONS OF A 3-DOF PROSTHETIC WRIST DEVICE

Neil M. Bajaj & Aaron M. Dollar

Yale University

ABSTRACT

This work discusses the hardware implementation details of a three degree of freedom prosthetic wrist device. To address some issues in current state of the art wrist design, we employ a hybrid mechanism architecture, consisting of a serial and parallel mechanism. This requires a combined view of both theoretical and practical physical considerations when designing such a device. We discuss kinematic analysis of the device, simulation methods and metrics that lead to informative physical implementation details, and actual physical implementations of the prosthetic wrist device. We show that these implementation characteristics can make different physical designs with the same nominal kinematics more suitable for different types of amputees.

INTRODUCTION

In activities of daily living, the human wrist may contribute as much to successful and timely completion of tasks as the fully dexterous, unaffected human hand [1]. Despite this, the design of prosthetic wrist devices has typically lagged behind that of prosthetic hands [2], though recently, more attention has been paid to wrist design, e.g. [3]–[7].

While the space of serial type wrist designs (i.e. joints placed in series) has been the standard for design of many wrist prostheses, borrowing ideas from parallel mechanism design has allowed for more freedom in wrist design. This additional freedom may be quite beneficial for prosthetic devices, for example, by allowing actuators to be placed proximal to the elbow (reducing elbow torque) and reducing the overall length of the wrist (making devices more suitable for amputees with long residual limbs). While these characteristics are obviously desirable in prosthetic wrist devices, designing parallel mechanisms to be incorporated into them can be quite challenging, particularly because of kinematics, actuation, and sensing considerations, combined with the small size scale of prosthetic devices.

We discuss the kinematic and physical design considerations in our proposed 3 DOF powered wrist prosthesis. The proposed device utilizes a hybrid (parallel and series) mechanism topology. Simulation of the kinematics is used to optimize the geometric parameters to improve the quality of motion and uniformity of torque production over the desired range of motion. Furthermore, we pay attention to key geometric metrics that would greatly affect the physical implementation of the prosthetic wrist, such as interference of the links of the mechanism on the structure, and range of motion of passive joints in the mechanism. We discuss further considerations that allow simple sensing schemes to be used, reducing the amount of hardware and cost of sensors required to provide closed loop feedback on the wrist.

METHODS

Kinematic Considerations

The proposed wrist design herein is composed of a hybrid mechanism, which is composed of a serial and parallel mechanism combined. Specifically, it is a $\underline{R}$U serial mechanism a 2DOF parallel U, $\underline{P}$RUR, $\underline{P}$SSR mechanism, where the underlined joints correspond to the actuated joints in each chain. The kinematic architecture and the methods of actuation can be seen in Fig. 1. The mechanism may be further broken into a flexion mechanism comprising the $\underline{P}$RUR portion, a deviation mechanism comprising the $\underline{P}$SSR portion, and the pronation mechanism comprising the $\underline{R}$U chain. This architecture creates spherical 3 DOF motion about the center of rotation, which is the passive Universal joint. As it can roll, a 2 DOF Universal joint becomes a 3 DOF Spherical joint, creating spherical motion emulating wrist motion.

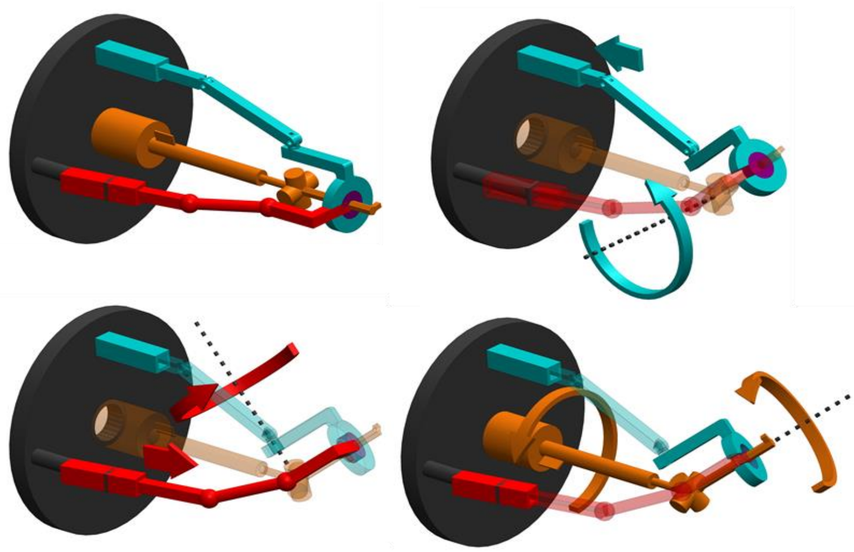

Figure 1: (Top left) kinematic representation of the 3 DOF prosthetic wrist device. (Top right) Actuating the PRUR flexion mechanism by actuating the blue prismatic joint. (Bottom left) Actuating the PSSR deviation mechanism about the indicated axis by actuating the red prismatic joint. (Bottom right) Actuating the RU pronation mechanism by actuating the pronation motor, creating roll about the indicated dashed axis.

This mechanism has a total of 7 geometric design parameters which may be changed to alter the motion and torque profiles of the overall mechanism. Moreover, an advantage of this mechanism architecture is that it partially decouples flexion, deviation, and pronation, meaning that only a single motor is largely responsible for each of the aforementioned wrist angles. This allows the flexion mechanism, deviation mechanism, and pronation mechanism to be simulated and optimized separately, and then the actual physical actuator may be changed to further alter the torque or speed characteristics, potentially to emulate anthropomorphic values. The kinematics are further described in [8].

To optimize each mechanism, we vary the geometric design parameters and then average the torque production capacity over a fixed range of motion (100º for flexion and deviation, 360º for pronation). Furthermore, we also track quantities relevant to the physical implementation. Namely, these are the minimum and maximum distance of any of the links from the central longitudinal axis of the wrist, and the range of motion required for each passive joint. Both of these factors are exceedingly important in the physical implementation and are often neglected in mechanism design, which hinders the transition from kinematic representation to physical prototype.

The minimum and maximum distance of the links from the central axis are important because they determine what the overall size scale of the mechanism will be when used in a wrist prosthesis. As this architecture can place actuators remotely, in a prosthetic device, they may be placed around the residual limb and socket. However, the mobile links of the mechanism must therefore not intersect or contact the socket during motion. The minimum distance metric therefore informs the designer of how much a mechanism with fixed geometric parameters would have to be scaled up to accommodate a residual limb of a fixed radius to avoid contact issues. The maximum distance is then related to this, as it then described how far from the central axis any link will need to be given this fixed radius. This tells a designer what the overall envelope of a particular design would be, and how "bulky" a mechanism would be compared to the residual limb.

Often, there will be loosely defined requirements on the minimum and maximum distance that would rule out some kinematic designs, especially when considering anthropomorphic proportions. These metrics are also closely related to the reversed workspace [9], and thus are related to the physical limitations of a wrist due to form factor.

Secondly, the range of motion of the passive joints is critical in being able to implement a particular design. Generally, passive joints (except revolute joints) have bounded ranges of motion that may often be exceeded in kinematic simulation. This is especially true for spherical and universal joints, though some work has been done to study and enlarge the range of motion of these joints [10]. Without considering their range of motion, during normal actuation of a mechanism, a passive joint will contact its physical joint limit, and large contact forces will arise on the links. Thus, it is critical that one constrain designs to those that have physically realizable passive joints.

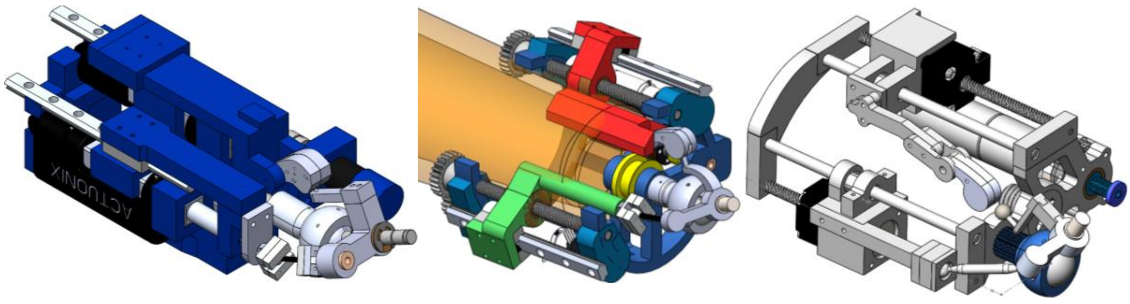

Figure 2: (Left) Minimum overall size implementation, compatible with a very short residual forearm or transhumeral prosthesis (no forearm remaining). (Center) Surrounding socket design, socket shown in transparency. (Right) Compromised design, compatible with short residual limb socket. Note that all wrists share the same size scale, with the central ball shown in each as a 1 inch (2.54cm) diameter sphere.

Physical Implementation

To verify and explore the importance and accuracy of the different metrics explored, we designed a series of physical representations of the optimal kinematic design. These designs correspond to 1) the minimal packaging required for the wrist, as if it would be placed directly on the end of the residual limb on the socket, 2) the design which would minimize the additional length required for the prosthetic wrist by placing as much of the wrist around an average prosthetic socket, and 3) a compromise design. Design 1 corresponds to the smallest maximum distance possible, design 2 corresponds to the smallest minimum distance that would accommodate a long residual limb socket, and design 3 corresponds to a short residual limb.

RESULTS

We present the physical implementations of the optimized design of the wrist device. All implementations share similar kinematic parameters with some overall scaling and translating of the actuators, though a comprehensive discussion of the allowable changes to facilitate comparison is too exhaustive for this paper. We also implemented the most size reducing actuator and drivetrain configuration possible (given a number of commercial and packaging constraints).

Figure 2 shows the three separate design implementations of the optimized prosthetic wrist prototype. The designs clearly show the effect of packaging requirements on the resulting physical design, where the actuators may be placed and what type of actuators and drivetrains must be used to accommodate the constraints on size. Design 1 allows for use of off the shelf linear servo motors, and package the pronation shaft about the center. The resulting design is discussed more in [5], but the overall size of 8.6cm length and a circumscribing radius of 4cm. Design 2 has an overall additional length on the end of the residual limb of 4.5cm, and a radius of 6.2cm. Moreover, the design can utilize a dc motor, gear reduction, and lead screw drive train for the actuated P joints, potentially increasing the mechanism torque and speed, though at the expense of complexity due to additional load bearing components. Finally, design 3 has an additional length on the residual limb of 5.6cm, with a cylindrical radius of 4.9cm. Note that this design could not fit linear servo motors or lead screw based drivetrains within its packaging limits while maintaining comparable torque to the other designs, so noncaptive linear stepper motors were used instead.

These designs also have a variety of different spherical joint range of motion requirements. We note that designs 1 and 2 require a large range of motion (130º conical range of motion), whereas design 3 requires a much smaller range of motion (70º conical range of motion). This affects the physical implementation of the spherical joints from the slot type design in design 1 and 2, and a symmetric standard ball and socket spherical joint in design 3. Moreover, the required passive hardware in design 1 and 2 may lead to much thicker components, affecting the necessary clearance to avoid collision.

To examine the validity of our simulations, the physical implementation of design 1 was tested for its torque and speed. Initial results are shown in figure 3. We characterize how quickly the wrist may actuate to different points in

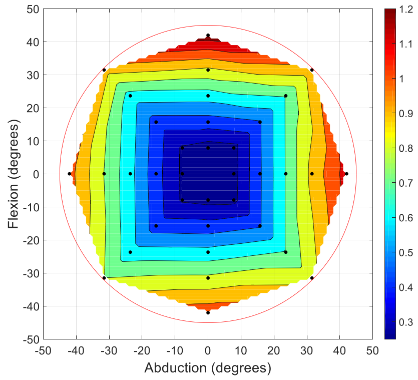

Figure 3: Time to actuate to the center speed over the range of motion of design 1. The distance between the contours is proportional to the maximum speed near that particular area, and is thus inversely proportional to the torque capacity of the mechanism.

its circumduction workspace. Through the use of electric motors, the speed of any point in the workspace is inversely proportional to the maximum torque at that point as well.

DISCUSSION

In this paper, we present practical implementation details of a 3DOF prosthetic wrist device, and show how under some constraints, this vastly affects the resulting physical implementation. We note that clearance and packaging constraints may influence the type of actuators used in a particular design, but more importantly, affect the type of residual limb geometry that is compatible with each design, potentially making different implementations of the same kinematics required for different amputees.

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A TRANSRADIAL MODULAR ADAPTABLE PLATFORM FOR EVALUATING PROSTHETIC FEEDBACK AND CONTROL STRATEGIES

Ben W. Hallworth[1], Ahmed W. Shehata[2], Michael R. Dawson[2], Florian Sperle[3], Mathilde Connan[3], Werner Friedl[3], Bernhard Vodermayer[3], Claudio Castellini[3], Jacqueline S. Hebert[2, 4], Patrick M. Pilarski[2]

[1]Department of Mechanical Engineering, University of Alberta, Canada. [2]Division of Physical Medicine and Rehabilitation, Department of Medicine, University of Alberta, Canada. [3]German Aerospace Center (DLR), Institute for Robotics and Mechatronics, Oberpfaffenhofen, Germany. [4]Glenrose Rehabilitation Hospital, Alberta Health Services, Canada

ABSTRACT

Novel multi-modal and closed-loop myoelectric control strategies may yield more robust, capable prostheses which improve quality of life for those affected by upper-limb loss. However, the translation of such systems from an experimental setting towards daily use by persons with limb loss is limited by the cost and complexity of assessing all the possible sensor and feedback configurations. The comparison of different control strategies is further complicated by the use of disparate prosthetic socket and simulated prosthesis designs across experiments. This study aims to address these issues through the development and preliminary assessment of a Modular-Adaptable Prosthetic Platform (MAPP) system for use in experimental control strategy evaluation. The MAPP system is compatible with a variety of commercially available control and feedback devices and can be used in experiments involving participants with either intact or amputated limbs. The modular design enables compatibility with novel devices and quick reconfiguration of components. We compared EMG and FMG data acquired with the MAPP system to a previously characterized transradial simulated prosthesis, using able-bodied subjects. The MAPP was shown to match or exceed the control accuracy achieved using a rigid simulated prosthesis, while providing the added benefits of modularity. This device shows promise as a research tool which can catalyze the deployment of advanced control strategies by enabling comprehensive and standardized assessment of control and feedback strategies.

INTRODUCTION

Recent developments in robotic prostheses have yielded many advancements including multi-articulated hands [1], [2], machine learning based controllers [3]–[5] and sensory feedback systems [6]–[8]. However, translating these improvements to wearable prosthetic devices remains challenging. Before translating these advancements to clinical use, thorough assessment and validation of the potential benefits are required. A significant bottleneck for assessment arises due to the tradeoff between experiment scale, representativeness of real-world conditions, and time/resource costs [9]. Numerous factors besides the control strategy itself, including end-effector loading, sweat, limb-position, and acceleration can affect the performance of a prosthetic system, and these conditions must be recreated during the experimental assessment to provide accurate insights into real-world performance [8], [10]. Simulating a realistic physical limb-socket interface within a participant- and control strategy-specific prosthesis requires a custom-designed and manufactured socket [10], [11], which is not easily adapted for various control and feedback systems.

An alternate strategy to custom-designing prosthetic sockets for testing persons with amputation is often pursued by having able-bodied persons wear a simulated prosthesis with or without an end-effector attached. Researchers have used various versions of simulated prostheses to investigate performance of commercial prosthetic hands [12], performance of novel control strategies [13], [14], kinematic movement trajectories when using prosthetic hands [15], and the effect of providing sensory feedback to users on performance in functional tasks [7]. There is, however, an incomplete understanding of how well results collected from these studies translate to daily use in a prosthesis by a person with limb loss. Furthermore, comparisons across studies are limited due to the disparate versions of the prostheses utilized. There is thus a need for a modular platform that accommodates multiple sensors and feedback systems and can be worn by both able-bodied persons and persons with amputations to facilitate these crucial comparisons. This study aims to address this gap through the design and assessment of an inexpensive and easy-to-use 3D-printed transradial Modular-Adaptable Prosthetic Platform (MAPP).

Figure 1: *Overview of the 3D-printable MAPP with a HANDI-hand attached to it* [2].

Table 1: *Design specifications for MAPP system*

Item	Design Specification	Achieved Specification
Length adjustability	10 – 40 cm	Achievable with multiple exterior panels
Fit intact limbs	Achieve Target	Target met
Prosthesis interface	Compatibility with iLimb, BeBionic, and HANDi Hand	Target met; expand modularity with new components
User input sensor integration	6 sites; compatible with commercially-available electrodes	10 sites; compatible with FSRs, MyoBock (Ottobock Inc.), and Bagnoli (Delsys, Inc.) electrodes
Context detection & sensory feedback	Accommodate 2 sensory-feedback modalities & IMU	Compatible with mechanotactile & vibrotactile feedback and IMU
Cost	$500	< $200
Fitting time	< 15 minutes	10 min initial fitting; 2-4 min re-donning
Socket weight	500 g	450 g
Shear/ axial load	2 kg	5 kg
Comfort	Comfortable over the course of an experiment (3 hrs)	Comfortable for 3 hrs (user-reported)
Sanitation	Non-porous, cleanable interface surface with limb	All contact surfaces lined with closed-cell neoprene

MATERIALS AND METHODS

Socket Design Requirements

Critical features were identified through consultation with prosthetists from the Glenrose Rehabilitation Hospital. Table 1 summarizes the design requirements and specifications for the developed socket. Unless otherwise stated, all components were 3D-printed using Ultimaker 2+ (Ultimaker BV) and Makerbot Replicator 2 (MakerBot Industries, LLC). Rigid components were printed using PLA and flexible components using Ninjaflex Cheetah filament (Ninjatek, Inc.). Figure 1 shows the design of the MAPP platform as a prosthetic socket for a person with transradial amputation. The developed socket consists of rigid panels supported by stainless steel M4 threaded rods with flexible cushions attached via Velcro® (Velcro BVBA). All panels are connected to a ring at the distal end of the socket.

Suspension

Suspension is achieved through radial compression generated by tightening the circumferential straps threaded through each rigid panel. Alternating regions of soft tissue compression and release are created by the cushions and spaces between them, distributed both radially and axially along the limb. This design choice improves translation of motion between bone and socket as described in [16].

Adaptability

To accommodate different limb lengths, the spacing between each 3D-printed panel can be adjusted and fixed by adjusting the position of the nuts embedded in each panel along the rods attached to the adjacent panel. A panel can also be removed entirely by unscrewing the rods which anchor it to the adjacent panel. This combination of modularity and adjustability enables the socket to accommodate residual limbs extending beyond 5 cm (the length of one panel) from the cubital fossa and up to 5 cm proximal to the wrist. Different limb thicknesses are accommodated by interchangeable inner rings with different diameters. As forearms are not cylindrical in nature, the channels in each panel through which the rod substructure passes are purposely made loose-fitting such that the slope between each panel can be adjusted. Furthermore, the interfacing cushions are made slightly compliant and convex such that they can match the profile of the limb surface without causing pinch points. When the circumferential straps are tightened, the socket profile is maintained due to opposing pressure exerted between each of the straps, cushion infill material, and limb surface (Figure 1). Able-bodied participants can be accommodated by replacing the connecting ring and distal support cushion with a hollow connecting ring. An optional hand mount can be screwed to that ring, thereby restraining the hand and fingers if isometric contractions are necessary. The hand mount, offset in the radial direction, directly fits with the Quick-Connect Wrist (Otto Bock, Inc.) to connect commercial end effectors. Custom 3D-printed adapters enable compatibility with other end-effectors.

Modularity and Socket Structure

The MAPP enables user input and sensory feedback devices to interface directly with a user's limb across a range of positions. Such devices can be embedded in each interior panel (Figure 2), providing a direct interface with the user's limb through which suspension loads are transferred. Rigid inserts provide a stable base for various actuators, which can be interchanged to accommodate other devices. Sensors can also be mounted in the spaces between regions with panels via the Velcro-backed circumferential straps. Velcro-backed modules prevent slip relative to the circumferential straps, and radial compression from the straps provides a stable interface with the user's limb. The interchangeable outer-panels add to the stability of this mounting method by securing the position of the circumferential straps relative to the rest of the socket structure with a Velcro-backed surface. Further, these outer panels provide an interchangeable platform for mounting devices (see Figure 1) on the socket's surface. A final method of modular device mounting is provided by the rails connecting the main panels. 3D-printed

Figure 2: *Exploded view of a) FSR and b) surface EMG electrode into panel system via removable inserts*.

Figure 3: *A participant wearing a) the Modular-Adaptable Prosthetic Platform as a simulated prosthesis and b) the orthotic splint.*

mounts can be threaded onto these rods providing a rigid platform which provides direct access to the user's limb via the spaces between exterior panels.

The interchangeable in-cushion sensor modules were designed to fit FSRs as described in [17]. Myobock 13E200 Electrodes (Ottobock Inc.) and Bagnoli Electrodes (Delsys, Inc.) were also made compatible with the initial prototype, enabling a mixed method of user-input detection. C2 and C3 vibrotactors (Engineering Acoustics Inc.) were similarly embedded into the interior cushion via interchangeable inserts, providing vibrotactile feedback in any cushion. 3D-printed mechanotactile tactor modules, the design of which is described in [8], were integrated into both the removable panels and substructure. The modularity of this socket system enables the integration of Inertial Measurement Units (IMU) (BNO055, Adafruit Industries) that could be used to detect forearm orientation and acceleration with respect to an inertial reference frame.

The structural rod segments were selected to support a 2 kg end effector load in both the transverse (ie. weight of 2 kg end load with residual limb parallel to ground) and axial (ie. 2 kg end load with residual limb perpendicular to ground). Using ASME Elliptic Failure Criteria and a life of at least 10,000 cycles of fully reversed loading, M4 rods were selected, leading to a minimum factor of safety of 2.5. The 3D-printed exterior panels were tested using both SolidWorks FEA (Dassault Systems, Inc.) and mechanical loading in the aforementioned configurations. These tests demonstrated that the overall minimum factor of safety was still limited by fatigue or bending of the rods; therefore, the socket system was capable of safely supporting up to a 2 kg end-effector or payload.

Socket Interface Validation Study

Participants: Eight able-bodied, right-handed, male participants (mean and standard deviation of age: 28.8 ± 8.2 years) volunteered to participate in this study. Written informed consent according to the University of Alberta Research Ethics Board (Pro00077893) and the German Aerospace Center's internal committee for personal data protection (DLR authorization 3.7.2017) was obtained.

Experimental setup: Participants conducted the experiment while wearing the developed MAPP (Figure 3a) and while using a version of an orthotic splint commonly used to simulate a prosthesis (Figure 3b). Participants were randomly assigned to start with one condition or the other. For each simulated prosthesis, a band of five evenly-spaced Myobock electrodes and a concentric band of five FSRs as described in [17] were placed on the participant's right forearm [18]. Signals from both bands were processed using the same hardware as [17], with a 3rd-order low-pass Butterworth filter and cut-off frequency of 1 Hz to remove high-frequency disturbances. Mean absolute value for each channel was extracted and used to train a linear-discriminant analysis (LDA) classifier, representative of commercially available classifier-based controllers [3]. An i-LIMB Ultra prosthetic hand was attached to simulate the effects of normal prosthesis loading on each socket (Figure 3). Participants were asked to match seven gestures (rest, index point, power grip, wrist flexion, wrist extension, forearm pronation, forearm supination) shown on a computer screen for two-second intervals, three times each.

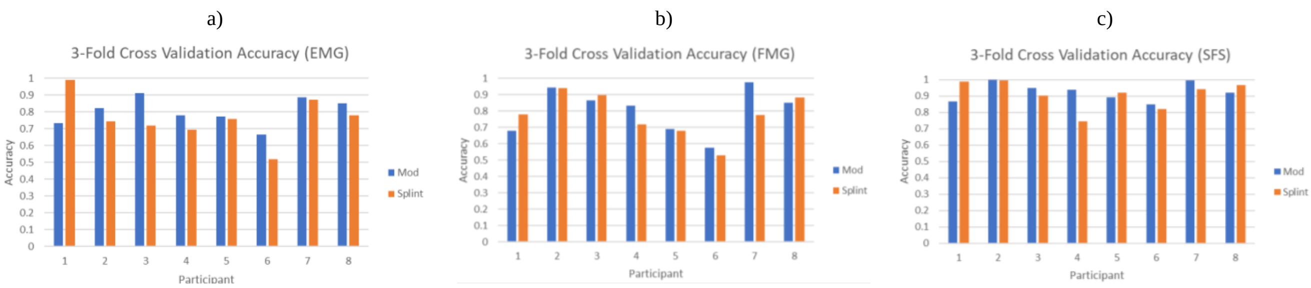

Figure 4. *Offline performance was assessed for each participant using a three-fold cross validation using a) EMG only, b) FMG only, and c) mixed-modality based on a sequential forward search (SFS)*

Data Acquisition: Offline performance was assessed for each participant using a three-fold cross validation (one for each repetition of a gesture). Assessment was performed using data from a) EMG only, b) FMG only, and c) mixed-modality based on a sequential forward search (SFS) to select the best-performance from 5 channels for each participant.

Results: Figure 4 shows that collecting data when using the MAPP enabled similar accuracy results as when using the orthotic splint across all sensor modalities.

DISCUSSION AND FUTURE WORK

Here, we developed a low-cost modular transradial socket system, which can accommodate multiple geometries of the forearm, along with multiple configurations of user-input, context detection, and sensory feedback devices. We tested the developed system with sEMG and FMG and a pattern recognition control strategy for seven gestures. Offline performance of participants using MAPP was similar to their performance when using the orthotic splint.

Future work will include comparison of online performance between the MAPP, orthotic splint, and socket systems. Using machine learning strategies to map input to action may reveal whether functional performance using a splint, or the MAPP provides a better prediction of clinical performance when deployed within a prosthetic socket. The effects of variables like end-effector loading, limb position, and acceleration are not well-characterized in control strategies. Therefore, paired assessment of the MAPP with a suction socket incorporating identical control strategies in different contexts may demonstrate the extent to which each platform captures these contextual changes. In conclusion, the cost time- and resource-savings, and flexibility to test a variety of novel prosthetic control strategies in a common platform, such as the one developed here, may accelerate the throughput of prosthetic control strategy validation.

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UPPER LIMB PROSTHESES – FUTURE PERSPECTIVES FOR BODY-POWERED PROSTHESES

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

Body powered upper-limb prostheses (bpp) have many advantages over EMG-controlled, electrically actuated ones (myo's), including mass, reliability, and proprioceptive feedback. Despite these advantages, bpp are rejected as often as myo's. Reasons mentioned include mass (despite being lower than myo's), and comfort (especially of the harness). In addition, recent research has shown the operating forces of bpp being too high. As a result the main advantage of bpp – feedback – is overshadowed, and the high operating forces negatively influence the comfort.

Current research at the Delft Institute of Prosthetics and Orthotics aims at improving the performance of upper-limb prostheses. First results show a promising future for prostheses controlled and/or powered by body movements, while satisfying the basic requirements for upper limb prostheses.

INTRODUCTION

For centuries mankind has tried to provide people with an arm defect with some kind of a replacement for the limb parts missing [1]. One of the oldest examples known, dating back to 330 B.C, is a prosthetic hand found on an Egyptian mummy. This device is a cosmetic hand prosthesis, i.e. without moving parts, primarily aiming at the restoration of the wearer's outward appearance. Dating from mediaeval times and some later ages, several examples of passive hands remain. Some of them with a moveable thumb only, some with the four fingers moving together in one finger block, and others with passive, individually adaptable, fingers. In these hands the thumb and finger configuration can be locked in a chosen position by the activation of a knob. A few examples are the famous hands of Götz von Berlichingen [2, 3] and the hands made by Ambroise Paré [1].

The beginning of the 19th century brings about a tentative start with actively operated prostheses. Harnessing gross movements of other body segments operates these prostheses. Hence, this type of prostheses is called body-powered (bpp). Examples include prostheses designed by Ballif in 1818 [2], by Van Peetersen in 1844 [2], and by the Count de Beaufort in 1860 [1]. Around 1900 the first attempts to power prostheses from an external energy source, most likely to relieve the user from the relatively high operating forces in body powered prostheses, can be seen. Examples include electrically powered prostheses [2, 4], or pneumatically powered ones [2, 5].
During WWII the idea of using myo-electric signals for the control of prostheses was conceived [6]. After extensive research and development myo-control evolved into the present day EMG-controlled, electrically actuated prostheses (myo's) and is still the subject for many researches to try and improve this control method.

At the Delft Institute of Prosthetics and Orthotics [DIPO] three basic requirements for upper limb prostheses were established: cosmesis, comfort, and control [7]. Judging bpp and myo's against these requirements it can be seen that bpp have many advantages over myo's, including mass, reliability, and proprioceptive feedback. Despite these advantages, bpp are rejected as often as myo's. Reasons mentioned include mass (despite being lower than myo's), and comfort (especially of the harness) [8]. Moreover, the functionality of myo's still lacks behind bpp (with the result of the Cybathlon 2016 as an example).
Recent research has shed even more light into why bpp are rejected: the operating forces are too high [9-11]. As a result the main advantage of bpp – feedback – is secluded, and the high operating forces negatively influence the comfort.

At the Delft Institute of Prosthetics and Orthotics (DIPO) current research aims at improving the performance of upper-limb prostheses.

METHODS

Within several ongoing projects DIPO tries to improve different aspects of upper-limb prostheses. Four of these projects will be highlighted here:

• Natural grasping
Within this project a body-powered voluntary closing hand prosthesis with adaptive fingers, a high pinch force to operating force ratio, and a low mass will be designed.

• Self-grasping hand
The goal of this study is 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.

• Haptic interface for prostheses control
This project aims to combine the advantages of externally powered prostheses (low operating effort, high pinch force) with the advantages of body-control (feedback). The idea is to measure movements of the body to control the aperture of the terminal device, and to measure pinch forces in the terminal device and feed them back to the body.

• Servo mechanisms
This project aims to enable prosthesis operation with low operating efforts. The envisioned servo mechanism uses pneumatic energy, as electro-mechanical servo mechanisms suffer from a high mass, and are sensitive for water and dirt.

RESULTS

The current status of the above mentioned project is discussed below.

• Natural grasping
A prototype hand was developed [12]. It has four adaptive, under-actuated fingers and a stationary thumb, Figure 1. The hand requires less energy (50-160%) of the user compared to current bpp-hands, while its mass is only 152 grams. Clinical test are ongoing.

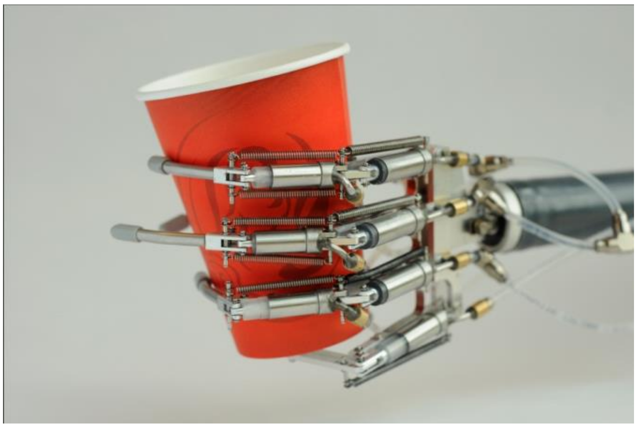

Figure 1 - The prototype of the Delft Cylinder Hand. It has four adaptive fingers actuated with two hydraulic cylinders in each finger, except for the little finger which has only one hydraulic actuator. The springs return the fingers to the open position at rest, and partly compensate for the counteracting forces of the cosmetic glove (not shown in the picture) as well. The cylinders in the hand receive the pressurized hydraulic fluid from a master cylinder incorporated in a shoulder harness.

• Self-grasping hand
Among the users of a hand prosthesis, about one-third uses a passive device. Nonetheless, little research is performed on improving passive hand prostheses [13]. At DIPO an innovative passive hand mechanism was designed. 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 hand, Figure 2, has a mass of 130 grams. A commercial version of this hand is almost ready for release.

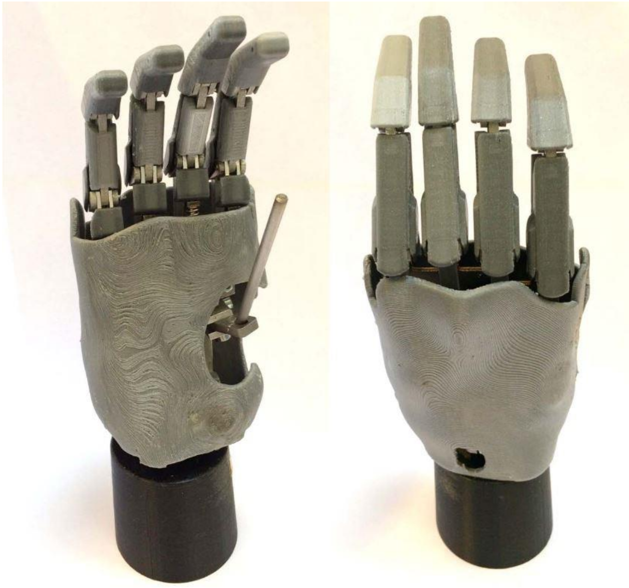

Figure 2 – The Self-grasping hand, shown without the cosmetic glove. In the right picture, the button to unlock the hand is visible on the dorsal side of the hand.

• Haptic interface for prostheses control
The designed interface utilizes skin anchors [14], Figure 3, connected by sensors and an actuator to record force/displacement and to provide feedback from sensors in the terminal device.

Figure 3 – The skin anchors placed on the body of a test subject. The cables are connected to the experimental set-up used verify the idea behind the haptic interface.

An experimental set-up, Figure 4, showed that the system indeed is able to provide input to the terminal device and gives proper feedback to the user [15]. Current activities include the design of a wearable actuator system.

Figure 4 – The experimental set-up. On the left the prosthetic simulator; in the middle and right part of the figure the master-slave unit is shown. Also visible are the cables and on the foreground, the skin anchors.

• Servo mechanisms
A hybrid system was designed that closes a voluntary closing terminal device by a Bowden cable as usual, and automatically activates a pneumatic servo as soon as an object is grasped. The output of the servo is proportional to the cable force, with a three-fold amplification.

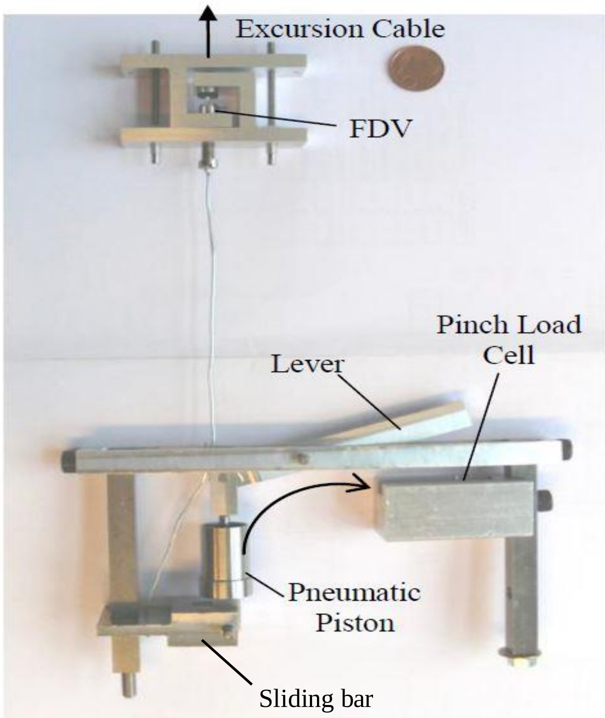

Figure 8 - An overview of the experimental setup. A cable (excursion cable) is connected to the force demand valve (FDV). The sliding bar will move when the excursion cable is pulled, this movement will cause the lever, which mimics a finger of the hand prosthesis, to rotate. Once the lever reaches the pinch load cell, representing the object to be grasped, the force in the excursion cable will rise. This increase in force will cause the FDV to start increasing its output pressure, which is connected to the pneumatic piston. This will cause the pneumatic piston to start applying force on the lever. The same force locks the sliding bar.

DISCUSSION AND CONCLUSION

The current projects at DIPO all show the future promises for upper-limb prostheses. The Delft Cylinder Hand is the first hand prosthesis that fulfils most requirements of the user: low mass, low operating effort, and proprioceptive feedback. The haptic interface shows a promising way of avoiding the harness, while maintaining the proprioceptive feedback. In combination with the pneumatic servo mechanism a prosthesis that combines body-control with a low operating effort comes within reach.

ACKNOWLEDGEMENTS

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VOICE RECOGNITION CONTROL OF A MULTI-ARTICULATING HAND FOR IMPROVED GRASP SELECTION

Benjamin McDonald[1]; Craig Kelly[1], Jen Johansson[1], Sam Lambrecht[1], Debra Latour, OTD,[2] Brianna Rozell[1], Todd Farrell, Ph.D.[1]

[1]Liberating Technologies, Inc. – A College Park Company

[2] Single Handed Solutions

ABSTRACT

About half of upper-limb (UL) amputees do not wear a prosthesis. This is, in part, related to an inability to take full functional advantage of the prosthesis. To help address this issue, we have developed the Voice Activated Prosthesis Interface (VAPI) to allow individuals to supplement their conventional control with voice commands. Specifically, this study targeted accessing multiple grip patterns in multi-articulating hands. Data from amputee test subjects is reported showing an improvement in the time to complete tasks, more accurate grip selection, and reduced frustration with the prosthesis when using the voice recognition technology compared to standard myoelectric control.

INTRODUCTION

It is generally agreed that only about half of upper-limb (UL) amputees wear a prosthesis [1,2]. This is often because the prosthesis does not return enough function for the burdens of weight, discomfort, non-cosmetic appearance, lack of durability, etc. [3]. One primary reason for the lack of prosthesis acceptance is the inability to control the device effectively. Difficulties with control result because multiple prosthetic joints are being controlled with a limited number of inputs from the user. The issue becomes even greater with more proximal amputation as there are even more joints to control with even fewer inputs available.

Current input options for UL amputees are limited and include switches, electromyographic (EMG) inputs from residual musculature, force sensitive resistors, linear transducers, etc. In addition, many amputees don't have the ability to use these inputs effectively (e.g., muscle atrophy can lead to unusable EMG signals). Also, conditions such as traumatic brain injury or other cognitive deficits can make it difficult to understand and produce reliable input signals. Even for proficient users, most current control strategies often require sequential control of the various system joints.

The lack of independent and intuitive control inputs also leads to existing complex prosthetic mechanisms being underutilized. For example, in recent years there have been substantial advancements in prosthetic mechanisms such as multi-articulating hands (Figure 1). These hands have the ability to produce dozens of different grip patterns that can be selected based upon the task being performed. However, grasp pattern selection can be complex and difficult to understand. Therefore, most users only utilize a maximum of four hand grasps due to the difficulty in reliably switching between grip patterns.

VOICE ACTIVATED PROSTHESIS INTERFACE

Upper limb amputees are looking for solutions that allow them to regain the function they lost after their amputation. To address this need, Liberating Technologies, Inc. (LTI) has developed the Voice Activated Prosthesis Interface (VAPI) controller which incorporates the ability for the amputee to use their voice to generate control signals for their prosthesis.

Speech is the most natural and highest bandwidth mode of communication for humans [4]. Therefore, we aim to augment users current control schemes with the addition of their voice as a new input modality. Using this approach, VAPI has the ability to access larger numbers of grip patterns within multi-articulating hands as well as fluidly perform tasks that require coordinated sequential movements of multiple prosthetic joints, such as opening a door.

Figure 1: *Prototype VAPI with iLimb Hand*

LTI has prototyped a fully embedded and stand-alone VAPI controller (Figure 1) to demonstrate feasibility of the concept. The current phase of research has been focused on developing three major components of the VAPI system: (1) the 'command interpreter' which interprets the commands through the voice recognition engine; (2) the 'command

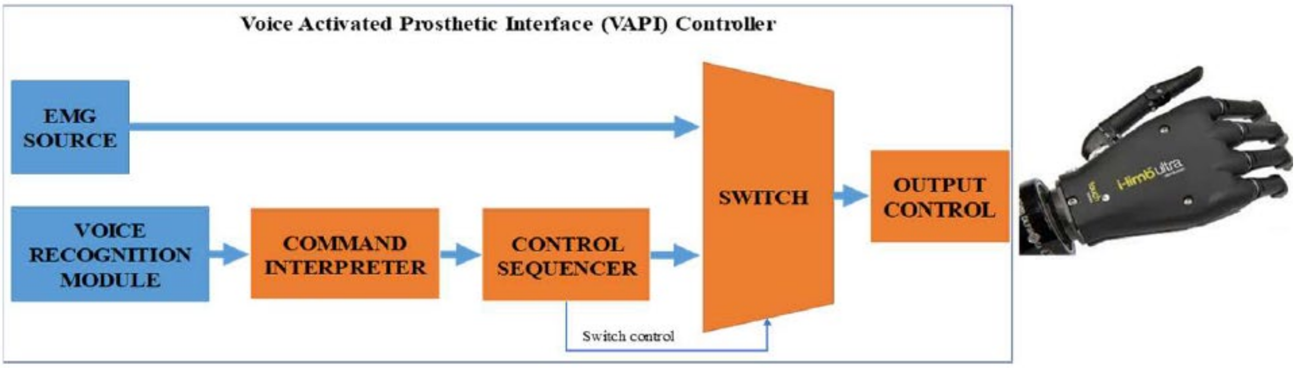

Figure 2: *Voice Activated Prosthesis Interface (VAPI) quick disconnect controller architecture.*

sequencer' which determines what control signals to generate based on the voice command; and (3) at the output controller to drive the terminal device (Figure 2).

As demonstrated in Figure 2 the VAPI is used in addition to the user's standard EMG signals, with the control sequencer determining if the EMG or voice inputs will control the terminal device. The VAPI system uses a trigger word to 'wake up' the voice recognition module (e.g., 'Alexa…') which then listens for the command word. This helps to reduce accidental activation of the prosthesis. After receiving the voice command, the VAPI produces the necessary command signal to elicit a grip change in the hand. Control is then relinquished back to the EMG sensors for the user to open or close the hand after the correct grasp pattern has been achieved.

VOICE COMMANDS AND RECOGNITION ACCURACY

With our partners at eSoftThings and Sensory, Inc., we developed a series of phonetically distinct command sets to test to determine which would produce the highest classification accuracy. Preliminary tests had six subjects perform five trials of each word in eleven different command sets. Figure 3 demonstrates that we were able to elicit up to 98% recognition accuracy for two of the command sets (#2&#8). The command set that was selected for future testing included the command words: "finger pinch," "power grip," "tripod," "key grip," "hand," and "wrist," with the latter two being used to toggle between hand and wrist control.

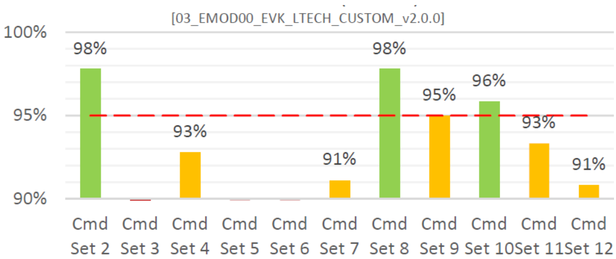

Figure 3: *Recognition accuracy results over 30 trials for each command set. Our goal was 95% (red dotted line).*

FUNCTIONAL OUTCOMES TESTING

<u>Methods</u>

We performed a set of standardized functional outcomes measures including the University of New Brunswick (UNB) Test of Prosthesis Function. While this test was originally intended for children, there has been shown to have acceptable reliability and preliminary evidence of validity for adults [5]. In addition to the UNB, we worked with our study Occupation Therapist (OT), Dr. Debra Latour, to develop a set of custom tasks that represent activities of daily living (ADL) where multiple grasp patterns may be useful. These included pouring and drinking, dressing tasks (put on sock, tie shoelaces, zip vest), turn doorknob, wrap a package and add written address label, etc.

Ultimately the user needs to generate the appropriate control signals to make the desired grip change. However, users will not always be able to switch into the desired grip pattern at the desired time. This could be due to imprecise muscle coordination (e.g., producing a double impulse when a triple impulse is required, not holding 'hold open' long enough, etc.), fatigue, misinterpreted voice commands, etc. Therefore, in addition to scoring the tests described above, we also tracked how often the subjects were not able to switch into the desired grip (i.e., 'Missed Grips').

Each subject completes the battery of tests either using EMG-only control or EMG with Voice Recognition (VR) control. Each subject was provided with an iLimb Ultra hand and VAPI for testing. In EMG-only mode, the hand was programmed to have four different mode switching commands used to access four different grip patterns within the iLimb (i.e., lateral, 3 jaw chuck closed, cylindrical, and precision pinch closed) via standard EMG switching commands including hold open, double pulse, triple pulse, and co-contraction. To ensure the length and weight of the prostheses were the same in both test conditions, the VAPI was installed, but disabled, during EMG-only control.

The subjects were trained on the VAPI and EMG switching and allowed to practice with each until they indicated they were comfortable with the control. Each subject then completed three trials of each task including both the UNB and custom tasks to simulate activities of daily living (ADLs).

Results

We have tested two subjects with limb loss thus far and testing is currently ongoing. We will have more subjects (both amputee and able-bodied) completed by the conclusion of the research funding in June of 2020. One subject with limb loss was an experienced two-site EMG user with a Touch Bionics iLimb multi-articulating hand. The other was a novice two-site EMG user with a Steeper beBionic multi-articulating hand. The experienced user was able to complete the full set of tasks three times. Due to fatigue, the novice user was unable to complete the full set of tasks. The novice user fatigued, in part, because the hand used for testing was significantly larger than their usual hand and the subject was unaccustomed to the additional weight.

UNB: Traditionally the UNB focuses on scoring spontaneity and skill. The measure of spontaneity defines a person's tendency and impulse to use their prosthesis effectively when attempting a two-handed task. In determining a person's level of skill, it may be evident that the person is able to perform the requested task but demonstrates the need for additional training or motivation to refine their abilities when using their prosthesis [6]. Scoring results from the UNB did not show a substantial improvement in spontaneity or skill for VR.

Timing: One of our original hypotheses was that voice recognition control would allow the user to complete their tasks faster. Our experienced user demonstrated that they were able to complete the tasks ***35% faster*** (13.3 seconds to 8.6 seconds) when using voice recognition (Figure 4).

Missed Grips: The experienced two-site myoelectric user that completed the full three rounds of testing was observed to have made 2.8 times more grip switching mistakes when using EMG-only control than when they used voice control (Table 1). These results were consistent across both the UNB and custom tasks. In addition, with EMG-only control, each missed grip would require an additional muscle exertion to achieve the desired grip.

Anecdotal Feedback: Missed grip changes were a substantial source of frustration for both control methods. Survey results demonstrated that both subjects preferred the voice control and had lower frustration levels with VR due to fewer grip transition mistakes. One user reported reduced exertion when using the voice control. Both subjects also reported frustration with the length and weight of the prototype system.

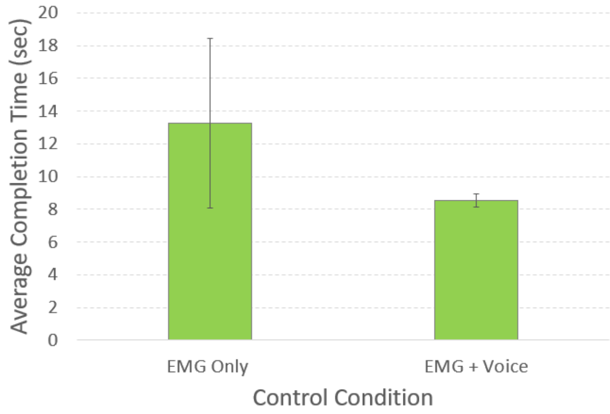

Figure 4: *Average task completion time with 95% confidence intervals with EMG Only grasp selection or EMG with voice control grasp switching.*

Table 1: Number of Missed Grips per control condition.

	Control Method		Ratio (EMG only/ EMG+Voice)
	EMG-only Control	***EMG + Voice Control***	
Missed Grips	51	18	2.8

DISCUSSION / CONCLUSION

Preliminary data indicates that voice recognition control of an upper limb prosthesis demonstrated more accurate multi-articulating hand grip selection than standard EMG-control methods. These data also indicated that it is possible to complete tasks more rapidly with voice control.

We believe that as individuals are able to easily and reliably access a greater number of grip patterns, they will be more likely to select the grip pattern that is ideal for the task at hand. With proper grip selection, it is likely that individuals will be able to reduce compensatory movements, which have been shown to lead to long term overuse injuries and joint damage [7].

It should be noted that there are other methods that can be used for selecting grip patterns that were not investigated in this study. These include the use of pattern recognition systems as well as the gesture control built into the iLimb Quantum. While these alternative methods are promising, there is a ceiling to the number of grasps that can be accessed through these methods. It has been reported anecdotally that pattern recognition can reliably access three to four grip patterns and the gesture control adds four patterns. We plan to compare voice recognition control to these methods in future trials.

FUTURE WORK

Testing of the VAPI system is currently ongoing with three additional amputee subjects and several able-bodied subjects to be completed by June 2020.

In addition to testing the current system, we are continuing to make further technical enhancements to the VAPI system with our current funding. One enhancement is to implement remote microphones, such as lapel or in-ear microphones, to detect and wirelessly transmit the voice commands to the VAPI for processing. This will move the microphone from its current location in the wrist, which has the potential to be interfered with if the individual were to choose to wear clothing such as a heavy, long-sleeved jacket. We will also investigate communicating directly with a multi-articulating hand itself over Bluetooth to be able to access an even larger number of grip patterns.

Finally, we are in the process of developing a new outcome measure specifically designed to assess the ability of individuals to access different grip patterns. We refer to this test at the Grip Switch Assessment (GSA). The GSA was inspired by the Box and Block Test, a commonly used measure to assess unilateral gross manual dexterity. The GSA was designed to measure a user's ability to efficiently switch between multi-articulating hand grips while manipulating simple objects. The assessment involves measuring the time it takes for a user to switch into the proper grip and carry a set of objects over a short obstacle (Figure 5). If the patient takes longer than 30 seconds to achieve the proper grip the test administrator will have the patient move onto the next item. This cut-off reduces the continued frustration of the patient and keeps the GSA trial time to under two minutes. The order of the objects is randomized with each trial.

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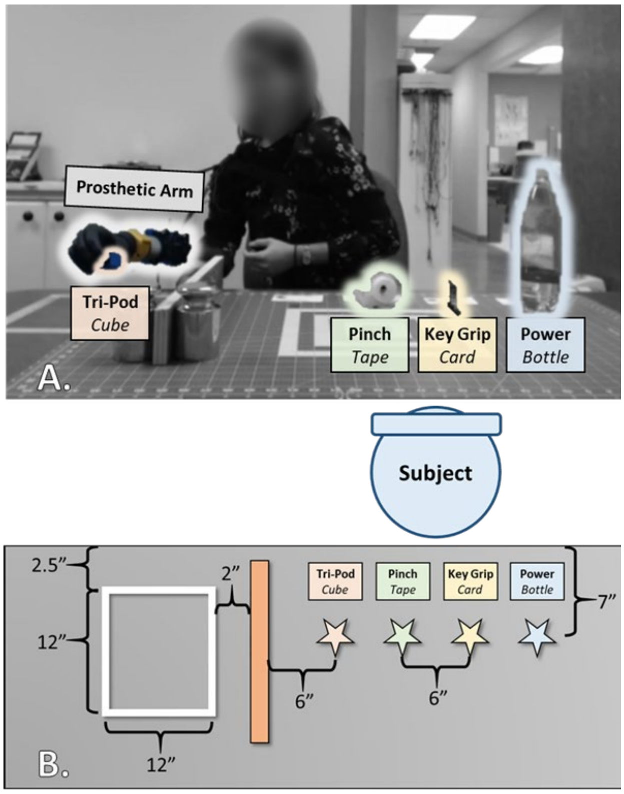

Figure 5: *A - An able-bodied participant manipulating the first object during a GSA trial. B - A diagram of the table arrangement to administer the GSA for a patient affected in the right arm.*

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Track: User Experience

A PLATFORM TO ASSESS BRAIN DYNAMICS REFLECTIVE OF COGNITIVE LOAD DURING PROSTHESIS USE

Oscar Ortiz[1], Usha Kuruganti[1], Daniel Blustein[2]

[1]*Faculty of Kinesiology, University of New Brunswick, Fredericton, NB, Canada.*
[2]*Department of Psychology and Neuroscience Program, Rhodes College, Memphis, TN, USA.*

ABSTRACT

Prosthetic hand operation often results in high levels of cognitive burden on the user which can lead to fatigue, frustration and device rejection. Some previous work that quantified this cognitive load relied on subjective questionnaires or distraction tasks. We have adapted a protocol capable of real-time, objective, non-distracting assessment of cognitive load for use with individuals controlling a prosthesis. Here we present this platform to assess cortical dynamics during prosthesis use. We describe a custom-built lightweight prosthesis simulator and an electroencephalography (EEG) assessment. We also present pilot work that shows how alpha inhibitory activity recorded with a wireless EEG system can be used to assess cognitive load.

INTRODUCTION

Efforts to improve upper-limb myoelectric prostheses often aim to provide a high degree of functionality to those living with limb-loss [1]. Despite technological advancement, these devices provide limited capabilities compared to intact limbs and impose a high cognitive load that results in fatigue and frustration [2], which can lead to device rejection [3]. Measurements to directly evaluate cognitive load are needed in order to further understand how efficient visuomotor behaviors develop during prosthesis use. For this, electroencephalography (EEG) is ideally suited as it allows the measurement of ongoing neural activity with high temporal resolution. Active processing in engaged and task-relevant areas of the brain is reflected by a suppression in the magnitude (power) of oscillations in the alpha range (8-12 Hz) [4], [5]. The development of skilled motor performance is characterized by the efficient allocation of processing resources to task-relevant areas of the brain [6]. Recently, this approach was used to demonstrate a decrease in alpha power detected across the scalp during prosthesis use compared to an anatomical hand, reflecting more conscious control [7]. Based on this work, we present a platform to assess brain dynamics during prosthesis use. The first section describes a customizable, lightweight myoelectric prosthesis simulator created for the platform. The second section describes the wireless EEG equipment and the analysis used in the platform. The project was approved by the Research Ethics Board of the University of New Brunswick (REB #2019-098) and all pilot testing was performed according to the REB guidelines. We conclude by showing pilot data of the alpha distribution on the cortex reflecting functional inhibition which can be indicative of high cognitive load.

METHODS AND PILOT RESULTS

Prosthesis simulator

A novel, custom built, lightweight (approx. 900 g) 3D-printed myoelectric prosthesis simulator was built (Figure 1). This device allows for people with intact limbs to control a prosthesis. The University of Alberta's Handi Hand [8] was mounted to a wrist brace with a medial offset, a position chosen to minimize the effect on modulating arm kinematics [9] and to reduce visual occlusion of the prosthesis [10]. Two electrodes (Myoware, Advancer Technologies) placed on the dorsal and ventral surfaces of the forearm record electromyographic (EMG) activity from wrist extensors and flexors to be used for hand control. Force sensitive resistors (Interlink Electronics®, CA USA) (FSRs) embedded in the fingertips of the index and thumb of the prosthetic hand detect pressure changes normal to the sensor that drive vibrating resonant motors providing haptic feedback to the user.

Control

Signals from the two EMG channels are amplified, high pass filtered at 20 Hz and notch filtered at 60 Hz. Signals are then rectified and integrated to drive a proportional open-close controller. Proportional control of the closing and opening velocity of the hand is done by mapping the maximal and minimal velocities to the maximal and minimal EMG activity recorded. To normalize the controller for each participant, they are asked to perform wrist flexion and extension maximal voluntary contractions (MVCs) for 5 seconds at the beginning of the session to determine the maximal amplitude for each of the electrodes. Similarly, the minimal activity for flexors and extensors is experimentally determined by recording the baseline EMG activity of each sensor during a period of 5 seconds while the arm is resting in the simulated prosthesis. The minimal activity is set to a value three standard deviations above the mean recorded activity to reduce unintentional activation of the channels.

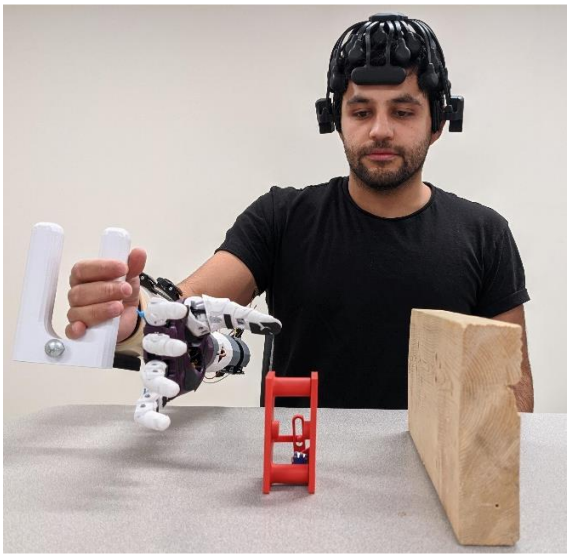

Figure 1. Experimental set-up displaying the custom prosthesis simulator and the dry-wireless EEG system. During experiments, the user's hand and arm are visually occluded.

Feedback

Changes in resistance captured by the FSRs at the fingertips control two haptic motor drivers (DRV265L, Adafruit Industries, New York, NY) that activate two corresponding linear resonant actuators (C10-100, Precision Microdrives, London, UK). These coin motors are in the inside lining of the forearm cuff and in direct contact with the skin of the forearm. The amplitude of the vibration of the haptic motors is mapped proportionally to the resistance change of the FSRs to represent the force detected at the fingertips. The magnitude of the minimally detectable vibration is determined individually for each participant and used as the lower edge of the mapping with the FSR signal.

EEG recordings

Cortical activity is recorded using EEG sampling at 1000 Hz. The electrodes are positioned on the head based on the standard 10/20 Channel system, with all referenced to the left and right earlobe. Data are transmitted wirelessly via Bluetooth from the cap directly to a PC and recorded using the software provided by the system manufacturer (Cognionics Data Acquisition, Version 3.6).

Blink and eye artifacts are removed using Principal Component Analysis and visual assessment [11]. EEG signals are then band-pass filtered from 0.1 to 100 Hz. Time-frequency decomposition of the signal is performed through short-time FFT on Hanning-tapered and zero-padded (up to 2000ms) overlapping segments (50% overlap) of 500 ms. These windows are recorded from 1000 ms before and after initial contact with the object to assess grasping force modulation (total time window of 2000 ms). Alpha power of EEG spectra has been previously used as a proxy to quantify functional inhibition of cortical areas [5], [7], [12], [13]. With this model, a greater level of alpha activity reflects a higher level of functional inhibition [5]. After the FFT transformation, power (μV^2) in the alpha range (8-12 Hz) is averaged across overlapping FFT segments for each channel and trial. Channels on the scalp are divided in 7 functional regions of interest (RoI); left temporal (T7), left central (C3), frontal (Fz), right central (C4), right temporal (T8), parietal (Pz) and occipital (O1, O2). Power is then averaged across these channels to yield values for each region. Finally, the values are divided by the average baseline value obtained during the resting state to obtain an index of change in activity from the resting state [14].

Using this method, we have been able to qualitatively identify high levels of alpha power reflective of functional inhibition of the occipital lobe during an eyes-closed recording. The occipital lobe is responsible for the processing of incoming visual information [15]. A sample recording from one participant is presented in Figure 2. This increase in alpha activity in posterior regions of the brain indicating low cortical activation has been well described since the late 1920's [15]. The wireless EEG setup presented here can identify alpha activity changes across the scalp.

DISCUSSION

A common goal in developing new myoelectric technology is to increase the clinical effectiveness of prostheses [3]. Despite advances in technology, most devices impose a high cognitive burden that can result in fatigue and frustration [2], and eventual prosthesis rejection [3], [16], [17]. Here, we present a platform to assess cognitive load during prosthesis use. The development of our prosthesis simulator facilitates experimentation with individuals not affected by limb-loss, allowing us to increase the statistical power of our studies. Furthermore, this system was manufactured using light-weight 3D printed parts, allowing for less constrained movements compared to previous simulators requiring suspension systems to offset the weight [10].

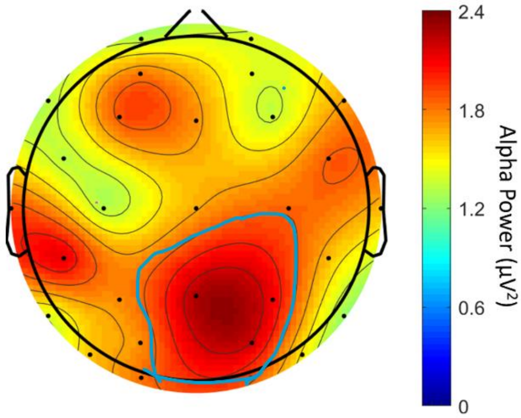

Figure 2. Sample alpha activity obtained during an eyes closed recording. Increased inhibitory alpha activity is present over the occipital lobe (outlined in blue), the area responsible for processing visual information [14].

Previous work has sought to assess cognitive load during prosthesis use using EEG [18], [19], however, only one previous study has attempted to directly evaluate the functional cortical dynamics using alpha level inhibition [7]. This work was able to demonstrate an overall reduction on alpha activity across the scalp during prosthesis use, indicating higher levels of cognitive load compared to the use of the anatomical hand. Based on this work, we present a platform aimed to help researchers and prosthesis developers investigate the effects of their prosthetic implementations on cognitive load. The advantage of our platform lies in the wireless EEG system utilized, as it does not restrict the movement of the user and avoids having large cable artifacts [20]. Furthermore, unlike the previous study using EEG to assess alpha activity [7], our protocol also includes a baseline normalization step, in which the relative differences in alpha activity between resting state and prosthesis use allows for the analysis of alpha changes exclusively due to prosthesis use, and allows for normalization across multiple assessment days [21].

From a practical perspective, it is important to understand how prosthesis users develop efficient control of a prosthesis. Adaptive learning processes rely on the engagement of appropriate mental resources during practice and performance [14], [22], [23], and high levels of cognitive load have been shown to hinder them [22], [24]. We hope to utilize this platform in the future to provide a method of assessing cognitive load during real time and move away from subjective or performance-based assessments of cognitive load as these are prone to subjective interpretations, distractions, and ceiling effects to tasks with high success rates [13]. Furthermore, EEG based assessments can provide insights about the cortical mechanisms responsible for the high levels of cognitive load, and drive evidence-based interventions on how to address them. Currently, we are conducting work using this EEG based approach to investigate the effects of adding augmented feedback on the cognitive load required to operate a myoelectric prosthesis, as augmented feedback could potentially reduce the visual attention and cognitive burden required to operate a prosthesis [18].

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USER-RELEVANT FACTORS THAT DETERMINE CHOICES FOR TYPE OF PROSTHESIS AND TYPE OF PROSTHESIS CONTROL

Nienke Kerver[1], Sacha van Twillert[2], Raoul M. Bongers[3], Corry K. van der Sluis[1]

[1] *University of Groningen, University Medical Center Groningen, Department of Rehabilitation Medicine, Groningen, the Netherlands*

[2] *University of Groningen, University Medical Center Groningen, Center of Expertise on Quality and Safety, Groningen, the Netherlands*

[3] *University of Groningen, University Medical Center Groningen, Department of Human Movement Sciences, Groningen, the Netherlands*

ABSTRACT

Background: We recently provided a comprehensive overview of factors that could determine prosthesis choice for persons with major unilateral upper limb defects by performing a qualitative meta-synthesis of literature combined with results from a focus group among end-users. However, this overview did not contain any user experiences about pattern recognition (PR) control. Therefore, the aim of the current paper was to validate the overview for PR controlled prostheses.

Methods & Materials: A literature search, in which we searched for qualitative contributions about PR controlled prostheses from the users' perspective, was performed. The relevant text in the results sections of retrieved papers was extracted and entered into Atlas.ti for a qualitative analysis. The coding framework was based on the overview of our recent meta-synthesis and focus group study. The overview consists of six main themes ('physical', 'activities and participation', 'mental', 'social', 'rehabilitation, costs and prosthetist services', and 'prosthesis related factors') and 86 subthemes.

Results: Three articles were included. Out of the 43 subthemes that were mentioned in the data, 41 were already included in the coding framework. The subthemes 'intuitiveness' and 'calibration' were added (both within the main theme 'prosthesis related factors'). Furthermore, results showed that PR control was experienced as intuitive, but also as unreliable, difficult and requiring extensive training and high mental effort.

Conclusion: An up-to-date overview with factors that could affect prosthesis choice, which consists of six main themes and 88 subthemes, that was also applicable to the choice for PR controlled prostheses was created. The up-to-date overview may help persons with upper limb defects to identify factors that really matter for them when selecting a prosthesis. However, since only three studies were included and only a limited literature search was performed, more qualitative studies about user experiences with PR controlled prostheses are needed to further validate the results of this paper.

INTRODUCTION

Considering the high rejection rates of upper limb prostheses, it is important to determine which prosthesis characteristics best suit the preferences of a user [1]. Therefore, we recently performed a study in which we identified user opinions about factors determining prosthesis choice for persons with major unilateral upper limb defects [2]. The study existed of two parts: a qualitative meta-synthesis of the literature and a validation of those results in a focus group with end-users [2]. Based on these results a well-arranged overview of 86 factors that could affect prosthesis choice was created [2]. Potential prosthesis users can use the overview, provided by the clinician, to identify what really matters to them. Users and clinicians can discuss those factors and select a prosthesis that best fits the needs of the user. However, one of the limitations of this study was that we did not include any user experiences with pattern recognition (PR) controlled prostheses [2]. Since prostheses with PR control have recently become commercially available, it would be beneficial for clinical practice to extent the overview for PR controlled prostheses.

In contrast to direct control (DC), which uses electromyography (EMG) signals of two muscles to control opening and closing, PR control uses algorithms that learn to recognize patterns from EMG of six to eight muscles [3,4]. In PR control, switching between different modes of the prosthesis by using a trigger signal (e.g. co-contraction) is not needed anymore. The appropriate grip is automatically selected based on the recognition of associated EMG patterns. In this way PR control aims to provide more intuitive control of the prosthesis. However, also disadvantages of PR controlled prostheses have been reported: they seem to be unreliable and require extensive training [5]. The aim of this paper was to validate the overview of factors contributing to prosthesis choice for PR controlled prostheses [2].

METHODS & MATERIALS

Coding framework

Our recently performed study existed of two parts [2]. In the first part a qualitative meta-synthesis using a ‘best-fit framework’ approach was performed [2]. For this meta-synthesis a systematic search of literature was done, in which studies were considered eligible if they contained qualitative content about adults with major unilateral upper limb defects experienced in using commercially available prostheses. Out of 6247 articles, 19 were included. In the second part of this study, results of the meta-synthesis were validated with end-users in a focus group [2]. The focus group included 11 persons with an upper limb defect, of which three used a standard myoelectric hand, three a multi-articulated myoelectric hand, one a standard and a multi-articulated myoelectric hand, two a cosmetic/passive hand and two did not use any prosthesis. The result of the study was a well-arranged overview of factors that could determine prosthesis choice for persons with major unilateral upper limb defects [2]. The overview contained 86 subthemes that were divided into six main themes: ‘physical’, ‘activities and participation’, ‘mental’, ‘social’, ‘rehabilitation, costs and prosthetist services’ and ‘prosthesis related factors’ [2]. Since we aimed to extend this overview for PR controlled prostheses, we applied the coding system used to create this overview as a coding framework in the current paper [2].

Data collection and analysis

A literature search, in which we searched for studies reporting on qualitative contributions about PR controlled prostheses from the users’ perspective, was performed (search date: 27-02-2020). PubMed was searched using the following search terms: ‘prosthesis’ AND 'upper limb' AND ‘qualitative’ AND 'pattern recognition'. Text was considered relevant if it was qualitative and described user experiences of persons with major unilateral upper limb defects with PR controlled prostheses. General information, such as participant demographics and analysis methods, were extracted and all relevant text in the results sections of the articles were extracted and entered into the Atlas.ti software. Relevant text included both quotes of participants and interpretations of the authors of the included studies. The data-extraction and analysis was performed by one coder (NK). If data did not fit within the existing themes and subthemes of the coding framework, new themes or subthemes were added. After a new theme or subtheme was added, the previously coded text was checked for the presence of this new theme or subtheme.

RESULTS

Study and participant characteristics

The electronic search resulted in three articles, which were included in this paper [5–7]. Those articles were not included in our recently performed meta-synthesis because in two studies non-commercially available prosthesis were used [5,7], in one study the focus on user opinions was not recognizable in the title or study aims [6], and one of the studies was published after the search we performed for the meta-synthesis [5]. A total of 24 adult participants were included in this synthesis (Table 1) [5,7]. In the study of Resnik et al. (2018) 12 adult participants used a PR controlled DEKA arm prototype in which the Coapt PR-control system was integrated with the DEKA-arm [7]. In the study of Franzke et al. (2019) four adult participants used a non-commercially available PR controlled prosthesis from Ottobock [5]. All 8 participants from the study of Hargrove et al. (2017) used a Boston digital elbow with a Motion Control wrist rotator and a single degree-of-freedom terminal device of their choice [6]. With exception of one participant, all participants were experienced with another prostheses [5–7].

Table 1: Summary of patient and study characteristics.

Study	Sample size	Gender	Origin of limb loss	Level of limb loss	Type of PR prosthesis	Other prosthesis	Country (ISO-code)	Data collection technique	Data analysis
Resnik et al. [7][A]	12	10 M; 2 F	1 CO 11 AA	10 TR; 2 TH[B]	3 EMG-PR-DEKA prototype 1; 6 EMG-PR-DEKA prototype 2; 3 both prototypes 1 and 2	11 personal prosthesis (type not specified); 1 none	USA	Open-ended questions in a survey and semi-structured interviews	Qualitative case series design with a constant comparison approach
Franzke et al. [5][C]	4	4 M	4 AA	4 TR	4 Michelangelo hands with non-commercially available PR control (Ottobock)	4 myoelectric prosthesis with DC control	AUT	Semi-structured interviews	Five-step framework approach
Hargrove et al. [6]	8	8 M	8 AA	8 TH[B]	8 Boston digital elbow with a motion control wrist rotator and a single degree-of-freedom terminal device of their choice[D]	8 myoelectric prosthesis (control type not specified)	USA	Open-ended question and an activities journal	Not clearly mentioned

[A] Study possibly also included persons with bilateral upper limb defects, however, this was not further described.
[B] All participants of those studies with limb loss at TH level also had TMR.
[C] Only participant demographics of the users with a PR controlled prostheses are shown in this table.
[D] Participants could choose between a powered split-hook (electric terminal device or electric Greifer terminal device) or a single degree-of-freedom hand.
ISO-code = country code assigned by the International Organisation for Standards; M = male; F = female; CO = congenital; AA = acquired amputation; TR = transradial; TH = transhumeral; TMR = targeted muscle reinnervation; EMG-PR-DEKA = a DEKA arm controlled by pattern recognition based on electromyography; DC = direct control; PR = pattern recognition control; USA = United States; AUT = Austria.

Findings

The data of current paper supported the six main themes of the coding framework. From the 86 subthemes of the coding framework, 41 were mentioned in the data. Most of these subthemes could be categorized within the main themes ‘prosthesis related factors’. Two new subthemes were added to the coding framework. The first subtheme was ‘calibration’ (main theme: ‘prosthesis related factors’), which was often experienced as inconsistent and unreliable. Since this issue was only mentioned in the study of Resnik et al. (2018), this might be explained by the prosthesis type with the PR control system that was used in this study [7]. Second, the subtheme ‘intuitiveness’ (main theme: ‘prosthesis related factors’) was added to the framework. PR was, if it worked well, often experienced as more intuitive compared to DC.

> *“Well, the PatRec [the pattern recognition control] surely is . . . with regard to how the control feels. . . more like it was before with the [intact] hand.” – Quote of a participant* [5].

On the other hand, regarding the subthemes ‘ease in controlling’ and ‘reliability’ (main theme both: ‘prosthesis related factors’), participants indicated that PR control was sometimes difficult and unreliable.

> *“. . .moving my arm in any way confuses it, I think, to where it thinks that I’m asking it to change the grip and it does when I don’t want it to.” – Quote of a participant* [7].

Additionally, with regard to the subthemes ‘prosthesis training’ (main theme: ‘rehabilitation, costs and prosthetist services’) and ‘mental effort needed to control’ (main theme: ‘mental’), participants said that extensive training and relatively high mental effort were needed for PR control.

> *“. . .it takes a lot more thought and a lot more training I feel, to, and not just like strength training and stuff, but just thinking of what muscles or what movements you want to make” – Quote of a participant* [7].

“First of all, pattern recognition requires a lot of training before it works properly.” – Quote of a participant [5].

DISCUSSION

This paper examined whether the overview of all factors that could determine prosthesis choice, that was created in our recent study based on results of a meta-synthesis and focus group, was also applicable to PR controlled prostheses [2]. Therefore, three studies that contained qualitative contributions about user experiences with PR controlled prostheses were synthesized using the overview of our recent study as a coding framework [5,7]. The subthemes ‘calibration’ and intuitiveness’ were added to the framework. This resulted in an up-to-date overview, which consists of six main themes and 88 subthemes, that was also applicable for the choice of PR controlled prostheses. Since PR controlled prostheses are already on the market, this up-to-date overview could be used in clinical practice to inform clinicians and prosthesis users about factors that may matter when selecting a prosthesis.

Results suggest that PR control was often experienced as more intuitive, but also as difficult to control, unreliable and requiring extensive training and high mental effort to control. These matters should be discussed between potential prosthesis users and clinicians when considering a PR controlled prosthesis. However, it should be noted that not all participants included in the synthesis of this paper used a commercially available PR controlled prosthesis. In addition, the participants of included studies were from a quite homogeneous sample (e.g. mainly males with an acquired amputation). Possibly, a different, more heterogeneous group of participants might have different experiences with PR controlled prostheses, and perhaps might have required further adjustments of the coding framework. Furthermore, the participants with an upper limb defect at transhumeral level had undergone targeted muscle reinnervation (TMR) [6,7], and in the study of Resnik et al. (2018) it was unclear whether participants with bilateral upper limb defects were included too [7], which may have influenced our results. Another limitation of this paper was that a limited search with only a few search terms in one database was performed. For this reason, we may have missed relevant information.

To conclude, this paper provides the first step in the understanding of factors that could influence the choice for a PR controlled prosthesis. The overview with factors that could affect prosthesis choice controlled by DC was updated for the use of PR controlled prostheses. However, since only three studies were included and a limited literature search was performed in this paper, more qualitative studies about user experiences with commercially available PR controlled prostheses are needed to further validate the created overview. We think that the updated overview of all factors that affect prosthesis choice, may help persons with upper limb defects to identify factors that really matter for them. Ultimately, we hope that this will facilitate a better match between user and prosthesis, resulting in a decrease of prosthesis abandonment.

ACKNOWLEDGEMENTS

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Track: Other

CASE STUDIES: FITTING PATIENTS WITH HEAVY DUTY RATCHETING MECHANICAL THUMB PROSTHESES FOR METOCARPOPHALANGEAL LEVEL AMPUTATIONS

[1]Ben Pulver, MS, [2]Mac Lang, CPO, FAAOP, [2]Rob Dodson, CPO, FAAOP, [1,3]Stephen Huddle, MS, [1,3]Richard Weir, PhD [1,4]Jacob Segil, PhD, [1]Levin Sliker, PhD

[1]Point Designs
[2]Arm Dynamics
[3]University of Colorado Denver | Anschutz Medical Campus
[4]University of Colorado Boulder

ABSTRACT

Thumb amputation presents a significant challenge for people due to the thumb's importance in creating stable functional grasps. Most thumb amputations are a result of trauma and most people with these amputations work in heavy manual labor occupations. The lack of many durable and functional prosthetic devices has caused many of these people to change or lose their jobs. This can lead to significant psychological and quality of life issues.

Here we present three different case studies of patients with metacarpophalangeal (MP) joint level thumb amputations being fit with a heavy duty ratcheting mechanical thumb prosthesis, the Point Thumb. The Point Thumb features anatomical flexion at both the MP and interphalangeal (IP) joints, a virtual MP joint center for better anatomical joint alignment, heavy duty metal construction, 10 different lockable positions, and the two methods of unlocking to allow for unilateral use. The first case is a patient with multiple digit amputations who desired to return to a manual labor job. The second case is a patient with an amputation of his dominant thumb who desired to improve effectiveness performing activities of daily living (ADLs). The third case is a patient with a left thumb amputation who desired to lift heavy objects to continue his hobbies and work. This patient had previous prosthesis experience and found the Point Thumb to be more functional than a cosmetic restoration or the TITAN Thumb. In all cases, the Point Thumb allowed patients to achieve their functional goals. These cases highlight the unique challenges present with thumb amputation and demonstrate the potential of the Point Thumb to provide users with a robust prosthetic thumb capable of handling heavy manual labor occupations.

INTRODUCTION

Approximately 500,000 people in the United States are currently living with an upper limb amputation [1]. About 92% of upper limb amputations are of the hand, finger, or thumb [1] and an estimated 45,000 new hand and finger amputations occur every year [2]. About 83% of these amputations are a result of trauma [1], [3]. The majority of these amputations are of fingers, 73%, with thumb amputations making up only 16% [4]. However, the loss of a thumb is far more significant than the loss of a finger; an amputation of the thumb at the MP joint leads to 40% impairment of hand function and 22% whole body impairment [5]. Additionally, the thumb is required to perform all but one of the most common grasps used to perform activities of daily living (ADLs) (Figure 1) [6]. Not only does the loss of a thumb create tremendous functional challenges, it can also create psychological challenges including depression, anxiety, social isolation, and low self-esteem [7], [8]. Despite the obvious importance of the thumb, recent studies have shown that the replantation rate for thumb amputations is declining [9] and patients are rarely fit with a prosthetic device of any kind [10].

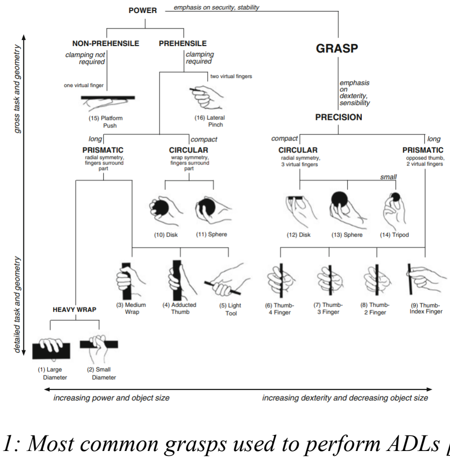

Figure 1: Most common grasps used to perform ADLs [6]

BACKGROUND

Clinical Significance

The thumb plays a critical role in hand function as it provides the primary source of opposition in nearly every functional grasp [6]. Thus, the nearly 74,000 people in the US with thumb amputations face significant functional challenges [1], [4]. For the thumb, an amputation at the MP joint leads to 22% of whole body impairment [5]. This degree of functional impairment can lead to job displacement as many of these amputations occur in heavy manual labor occupations which can no longer be performed after the amputation (Figure 2).

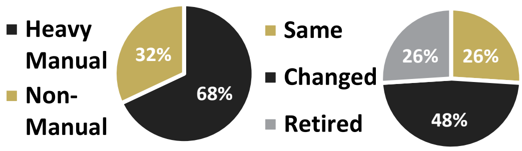

Figure 2: (Top) Work performed prior to partial hand amputation. (Bottom) Job status after receiving partial hand amputation [11]

Prosthetic Options

There are several prosthetic options currently available for people with thumb amputations. In general, they can be sorted into four categories: cosmetic, body-powered, passive/positional, and externally powered (Figure 3).

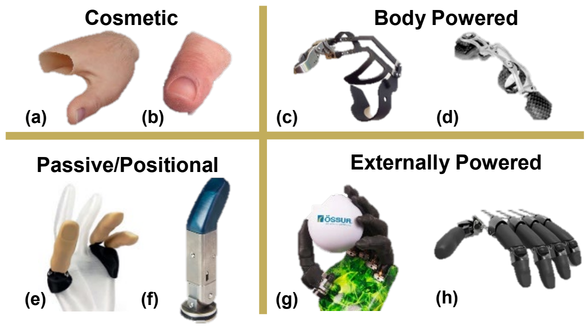

Figure 3: Overview of prosthetic solutions for thumb amputations. (a) custom silicone thumb (stamos and braun prothesenwerk) (b) livingskin™ (Ossur) (c) X-Thumb (Didrick Medical) (d) ThumbDriver (Naked Prosthetics) (e) VINCENTpartial passive (Vincent Systems) (f) TITAN Thumb (Partial Hand Solutions) (g) i-Digits Access (Ossur) (h) VINCENTpartial active (Vincent Systems)

Cosmetic devices, such as livingskin™ (Ossur), are mostly an aesthetic option and provide limited functionality. Body-powered devices, such as the X-Finger (Didrick Medical) and the ThumbDriver (Naked Prosthetics), are more functional by providing active flexion and opposition. These devices are limited, however, by their reliance on a custom fit and limited grip force. Passive/positional devices, such as the VINCENTpartial passive (Vincent Systems) and TITAN Thumb (Partial Hand Solutions), provide adjustable flexion and opposition so are generally more functional than cosmetic solutions. These devices, however, often require the use of the user's contralateral hand to position the device. Externally powered devices, such as the VINCENTpartial active (Vincent Systems) and i-Digits Access (Ossur), are controlled using myoelectric signals and provide active flexion, manual or active adduction, and active opposition. Durability and intuitive control systems are generally a challenge with these types of devices.

Table 1 provides a comparison of the different prosthetic options available in terms of their range of motion. The impairment values are calculated using the American Medical Association (AMA) guide for evaluating upper extremity impairment [5]. This comparison does not factor in issues like loss of sensation, device durability, and device ease of use, all of which have a significant role in device adoption and retention. Even so, this shows that large functional gains can be made by simply including flexion at one or two joints.

Table 1: Thumb prosthesis functional comparison from the perspective of digit and hand impairment remaining after fitting the prosthesis.

Prosthesis	Examples	Impairment*	
		Digit	Hand
No Device	---	100%	40%
Static Opposition Post	livingskin™	55%	22%
MP Flexion	TITAN Thumb	37%	15%
MP and IP Flexion	Point Thumb	31%	12%
MP Flexion and Radial Abduction	VINCENTpartial passive	27%	11%
MP Flexion and Adduction	i-Digits Access[1] VINCENTpartial active[2]	17%	7%

[1]Adduction is passive, [2]Adduction is active
**Does not include impairment due to lack of sensory information*

As durability is a key issue for people desiring to return to work in heavy manual labor jobs, body-powered and passive/positional devices are generally preferred. Despite this preference, there are still limited options for heavy-duty devices and thus new devices must be developed.

Point Thumb

The Point Thumb, by Point Designs, is a new heavy-duty passive/positional device with 10 different lockable positions in flexion and two degrees of freedom (DoFs) (Figure 4). It is the only device that features motion at the IP joint to achieve anatomical flexion as well as the only device to feature a virtual MP joint center to achieve anatomical joint alignment. With two methods of unlocking the ratchet mechanism, it is also able to be used unilaterally.

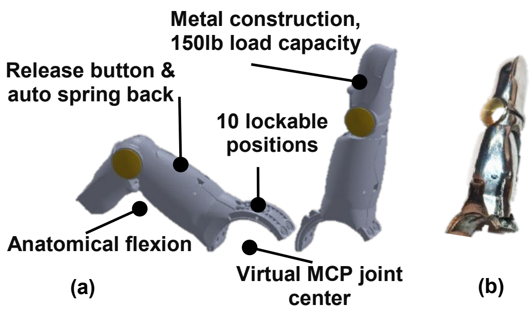

Figure 4: (a) Rendering of Point Thumb prototype with design features highlighted. (b) Physical Point Thumb prototype

CASE STUDY 1

Presentation

The first patient is a 49-year-old male who sustained a workplace injury resulting in a partial hand amputation of the left 1st-3rd digits at MP joint and 4th digit distal to IP joint. At the time of the initial clinical evaluation he and his wife were caring for 7 foster children including 2 infants. He has seasonal work as a firefighter which he aims to return to. He is also considering returning to his previous job as a laborer which requires handling tools, lumber and heavy bags of supplies.

Figure 5: (Left) Patient's presentation and prosthesis with Point Thumb, two Point Digits, and one Point Partial. (Right) Patient lifting a weight with prosthesis.

Treatment

Due to the ruggedness of his occupational goals, passively positionable digits were recommended to improve grasp security. The intended use of the prosthesis was for work and ADLs including his hobby of logging. Externally powered options were contraindicated for his reported goals. The Point Thumb was considered a good option due to its robustness and ability to flex at the IP joint, which in this case was critical for achieving opposition with digits 1 and 2.

The patient was fit with a partial hand custom high temperature vulcanized (HTV) silicone socket and carbon fiber frame. The Point Thumb was used for the 1st digit and two full length Point Digits (Point Designs) were used for the 2nd and 3rd digits. Additionally, a partial finger prosthesis, the Point Partial (Point Designs), was used for the 4th digit by creating a separate custom HTV thimble style socket.

Outcome

The patient was able to securely hold long handled tools and cylindrical items. Pinch grip was made possible by the attachment of the Point Thumb mounting bracket to the silicone socket rather than the carbon frame. This flexibility allowed for some adduction to improve opposition, particularly active opposition between the Point Thumb and the 4th and 5th digits.

The patient adapted to use of the prosthesis quickly. Within one month the patient reported using the device to assist in chainsaw operation as well as use of an axe. He reported wear of the prosthesis up to 12 hours per day without issue but with an average of 4 to 6 hours.

The Disabilities of the Arm, Shoulder, and Hand (DASH) standardized outcome measure was used to assess prosthesis effectiveness. The patient experienced a reduction in DASH score from 22 to 15, which while not meeting the minimum clinically important difference demonstrates important functional gains from the Point Thumb.

CASE STUDY 2

Presentation

The second patient is a 36-year-old male who sustained a right dominant thumb amputation secondary to a workplace accident. He previously worked in corrections and at the time of the initial clinical evaluation was considering alternate career options. He did, however, express a desire to return to his prior employer in some capacity and for some time.

While recovering from his injury, he is the primary caregiver for his children, while his wife works full time. He has difficulty with numerous ADLs given decreased ability to pinch and grasp with his previously dominant hand. Measurements taken during hand therapy indicated an 85% reduction in hand strength of his dominant hand compared to his non-dominant hand.

Figure 6: Patient's socket with Point Thumb prothesis

Treatment

The patient's goals dictated a digit for opposition that would be durable and very strong. His occupation necessitated a variety of thumb positions to provide pinch of flat lumber as well as grasp of round handles and tools. This requirement indicated he would benefit from the Point Thumb as it has motion at both the MP and IP joints.

Outcome

The patient was fit with a partial hand custom HTV silicone socket and carbon fiber frame. The Point Thumb was integrated rigidly into the carbon frame with alignment allowing for precision pinch, tripod pinch, as well as cylindrical and spherical grasps. More quantitative outcome measures will be reported after the patient has used the new device for an extended period.

CASE STUDY 3

Presentation

The third trial patient is a 57-year-old male who sustained a workplace injury resulting in the MP level amputation of the left thumb (Figure 7). At the time of the initial clinical evaluation he was working in a construction environment, mainly in carpentry. His main functional goal was the ability to grasp objects such as tools and materials such as lumber to perform his daily tasks at work, continue working on cars as a hobby, and perform ADLs at home.

Figure 7: (Left) Patient's presentation. (Right) Patient using Point Thumb to hold a spray bottle

Treatment

The patient was initially fit with a custom silicone restoration and a passively positional thumb, the TITAN Thumb, attached to a dynamic muscle contoured interface. An externally powered thumb was contraindicated due to the patient's bulbus distal presentation as well as a dirty and possibly wet working environment.

The patient found that the cosmetic restoration did not allow him to grasp heavy objects. While the TITAN Thumb gave the patient increased ability to grasp heavy objects, the patient found the need to use his contralateral hand to unlock it unacceptable. The Point Thumb was then fit as a replacement to the TITAN Thumb and found to correct this issue by allowing unilateral use.

The patient was ultimately fit with a partial hand custom HTV silicone socket and carbon fiber frame. The Point Thumb was integrated into a carbon fiber thumb cap that was glued to the HTV silicone underneath and allowed for grasp of both large and small objects.

Outcome

The patient reported increased satisfaction with the Point Thumb due to the novel spring back mechanism. This trial fitting was very recent and thus the collection of standardized outcome measures data is ongoing. Further results will be reported after the patient has used the new device for an extended period.

CONCLUSION

Thumb amputations present a variety of complicated functional, psychological, and occupational challenges. Most people with thumb amputations work in heavy manual labor occupations and the lack of robust prosthetic options up to this point prevents many of them from returning to work. The Point Thumb is a new robust passively positionable ratcheting prosthetic thumb with flexion at the MP and IP joints designed for use in heavy-duty work environments. The three case studies presented here illustrate the complexity of thumb amputation cases and demonstrate the viability of the Point Thumb as a robust prosthetic thumb for heavy manual labor occupations. In all cases, use of the Point Thumb allowed patients to achieve their functional goals, ranging from using a chainsaw to carrying lumber. These positive early trial fittings indicate that the Point Thumb has strong potential and warrants further study.

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CHANGES IN TECHNOLOGIES AND MEANINGS OF UPPER LIMB PROSTHETICS: PART I - FROM ANCIENT EGYPT TO EARLY MODERN EUROPE

David Foord

Assistant Professor, Faculty of Management, University of New Brunswick

ABSTRACT

This paper is the first-of-a-three-part series that examines changes in technologies and meanings of upper limb prosthetics from ancient Egypt to the present. Contemporary design of powered upper-limbs shares a number of continuities with older methods of limb making. Both cosmetic and functional prostheses have been in production for at least the last 3,000 years. Wars have long been a spur to technological innovation in artificial limbs. Prostheses making has sought to return soldier-amputees to combat, whether on horseback or in tanks. Technological innovation in other fields has provided materials for improvement in artificial limbs, such as the replacement of wood by iron, iron by steel, or plastic by composites. Since the early modern period, the development of new artificial limbs has been mistaken as a specialty of medical doctors. The making of artificial limbs has since at least Ancient Egypt been as much about technological innovation as the creation of new meanings of prostheses, whether the design of arms for the underworld of the Duat or hands as industrial tools.

INTRODUCTION

The management scholar Roberto Verganti has both combined and expanded our definitions of *innovation* and *design* through the concept of design-driven innovation. In his definition of *design*, it not only includes design as a form of product, e.g. bringing visions of beauty to products, or design as creative problem solving, e.g. human centered-design anchored in user discovery, but a third meaning of design as "making sense of things [1]." In this third meaning, design contributes new meanings, whether combined with radical or incremental technological innovation. In this paper we examine how multiple meanings of powered upper prostheses have emerged in the design of these limbs, including "prosthesis as afterlife limb", "prothesis as battle tool" and "prothesis as natural limb".

ANCIENT UPPER LIMB PROSTHESES

Recent studies are revising accounts of ancient Greece and Rome as the source of the first prostheses, and medieval Europe as a dark age of heavy, crude prostheses for battle and hiding amputations [2]. There is evidence from ancient Egypt that artificial body parts were used to reinstate the physical body for its reanimation and continued existence in the next life. As well, functional prostheses were designed and used for mobility [3]. Artificial big toes found with Egyptian mummies and dated to the 11th to 7th centuries BCE show both a realistic appearance and functional performance in contemporary walking tests with replicas [4]. The inference is that a "nascent prosthetic science may have been emerging in the Nile Valley as early as 950 to 710 BC", perhaps demonstrating ancient Egyptian knowledge of human anatomy [5]. Upper-limb prostheses, in contrast, were rarely used in ancient Egypt, and then only by the rich. The oldest Egyptian cosmetic hand found on a mummy dates to 2000 BCE [6]. Another cosmetic hand has been dated to 200 BCE [7].

Beyond Egypt, there is evidence of prostheses in ancient Greece, Rome, Peru and China. In Greek mythology, the god Hephaestus, sometimes shown with shriveled foot, is skilled in the technical arts and making of prostheses [8] Other written records include Herodotus' description of the wooden foot and an iron arm prosthesis of the 3rd century BCE Roman general, Marcus Sergius Silus [9].

Although archaeological evidence for amputations before 1000 is uncommon, there are reports of artificial limbs during the first millennium. Artificial feet have been found in burials in Bonaduz, Switzerland from the 5th to 7th century and Griesheim, France, dated to the 7th to 8th century. A male from Longobard, Italy during the 6th to 8th century was found with a forelimb amputation and prosthetic device [10]. The historian Reed Benhamou in a *Technology and Culture* article on the history of the artificial limb in preindustrial France wrote that "Artificial hands capable of at least the palmar pinch required to hold a pen may have been made as early as the 10th century [11]."

EARLY MODERN LIMB PROSTHESES

As with amputations before the year 1000, there is little historical, archaeological or iconographic evidence of artificial limbs before the 16th century [12]. Most amputations may have been lethal, as techniques to stop extensive bleeding only became widely known in the 17th century. As well, there was a lack of knowledge of how to avoid infection. In the sixteenth century, doctors cauterized gunshot wounds with boiling oil. It was only after the French army barber-surgeon Ambroise Paré (1510–90) ran out of oil in a battle that he began using ointments and dressing as an alternative. Seeing improvement in his patients, Paré discovered that ligation of blood vessels controlled bleeding during amputation [13]. However, even in those rare cases in which severe upper limb trauma did not result in deadly hemorrhage and infection, only the wealthy could afford customized prosthetic upper limbs.

The growing affluence in the early modern period is evidenced by the holdings of artificial upper limbs in the London Science Museum. It has five upper, artificial limbs dating from the 16th to 18th centuries [14]. The lack of lower limbs in the collection, despite being more numerous than upper limbs both then and now, is likely attributable to the use of wood and leather in artificial legs (excepting for knights on horseback), versus iron or steel for upper limbs. Among the most famous prosthesis of the early modern period is the iron hand of German knight Götz von Berlichingen (1480-1562). An artisan made the prosthesis battle armour for him after he lost his hand during the Siege of Landshut (circa 1505) in Bavaria. It featured five digits that were capable of a fingertip pinch, could be flexed and locked so he could to hold reins, grasp and swing his sword and return to battle with disability concealed. The iron upper limb of a Turkish pirate, Horuk Barbarossa, was discovered in an Alsatian tomb dated to 1564. He lost his hand in the Battle of Bugia (circa 1517) against Spain, and, like Götz von Berlichingen, received an iron replacement so he could fight again in battle. The Barbarossa upper limb also featured a movable wrist joint, elbow joint and fingers. Following a similar theme, Duke Christian of Brunswick lost his left hand in the Battle of Fleury (circa 1622) and received an iron hand from a Dutch craftsman. The only account of a non-combative hand prosthesis from the period comes from outside of Europe. It is attributed to the Italian surgeon Giovanni Tommaso Minadoi, who in 1512 "while travelling in Asia recorded observations of an upper limb amputee who was able to remove his hat, open his purse, and sign his name [15]."

According to numerous accounts, the first prosthetic device "that demonstrated a sound understanding of basic biomechanical functions" was designed by Paré [16]. It was a mechanical hand operated by catches and springs. It sought to copy with metal the motion and appearance of the missing hand, as well as offer beauty and ornament to wealthy patrons. In some reports, the hand met with limited success. In others, it was heavy but "successfully restored a knight's ability to hold a shield or weapon in battle" and "were carefully crafted with the shape and appearance of human hands, rather than simply inanimate tools to hold objects [17]." It was not, however, designed by Paré, but obtained from a locksmith living in Paris, known as "le petit Lorrain." [18] According to Heidi Hausse, it provides evidence of "ongoing practices of creating prosthetic technology in this period" within the domain of locksmiths, gunsmiths, clockmakers, and armorers in the early modern period. Neither Paré nor the surgeons who read his *Oeuvres* could fabricate the prosthesis or understand design, as opposed to the master craftsmen, who were capable of both. Rather it was the surgeons of wealthy patients who could afford these luxury items containing expensive materials and new technology.

A similar division between surgeons and master craftsmen is seen during the English Civil War (1642–51). "The Hospital of the Savoy in London treated amputee soldiers whose prosthetic requirements survive in credit bills submitted by William Bradley, hospital carpenter, indicating he supplied wooden legs and their attachments, made repairs, and provided crutches of various lengths [19]." This also presents evidence of the beginning of the institutionalization of prosthetic device design and manufacturing within hospitals.

EIGHTEENTH CENTURY LIMB PROSTHESES

The eighteenth century saw the refinement of early modern upper limbs, the expansion of craft businesses into advertising, the beginning rehabilitation programs in hospitals and new ideas of politeness that led to expressions of uneasiness with visible disability. The refinements included body powered upper limbs. In 1732, the Académie Royale des Sciences reviewed a below elbow artificial arm designed by a clockmaker named Kreigseissen. It was made from sheets of copper and had joints at the wrist and at the first and second knuckles. These joints also offered lateral movement of the thumb to accomplish a palmar pinch were accomplished with pulleys activated by bending the elbow. There was also reduction in the weight of upper limbs. The Götz von Berlichingen artificial hand weighed about 1.5 kilograms, a little less than the average male hand and forearm. By 1792, an artificial arm made in Switzerland had less than a third of the weight due to the use of steel instead of iron for the springs, barrel casings, cables, and triggers, papier-mâché and parchment for the forearm, and cork for the wrist and hand rather than metal or wood. As evidence of the continuing role of master craftsman in the design of prostheses, Reed Benhamou wrote that "It is no coincidence that these are the same materials used in 18th-century automatons, for several of the clockmakers, locksmiths, and *mécaniciens* who used the lightweight materials and miniaturization techniques required by these devices also produced artificial limbs. Indeed, some 18th-century prostheses may be called spin-offs of technology [20]."

Shaping the response to wartime limb loss were Enlightenment values and a new culture of politeness. In early the early eighteenth century visible "'deformity' made others uneasy and threatened the virtuous social interaction or 'conversation' that lay at the core of notions of politeness [21]." Moreover, in Britain, many "condemned the use of prosthetic technologies as deceitful, prideful and impious [22]." By the late eighteenth century, prostheses came to be associated with Enlightenment visions of scientific discovery of the mechanics of the human body and technological progress in replicating nature. In this cultural shift the association of prosthesis with the sin of pride and Puritan ideals of unadorned purity was reformulated as artificial improvement. Prosthesis use came to be seen as restoring wholeness and normalcy and practicing polite behaviour in putting others at social ease. In this way prostheses users were meeting their moral duties of a sociable society. With this context the advertisement of prostheses as devices of scientific improvement further influenced the growth of this idea of politeness, with as much emphasis on agreeableness, decorum and taste, as it did with the medical marketplace. Toward the end of the 18th century, the French Revolutionary Wars (1792-1802) led to new opportunities for sales of protheses. In addition to tailoring prostheses for individual amputees, "craftsmen solicited endorsements, advertised their products in the popular press, and, in general, attempted to sell prostheses as they might any other commodity [23]."

These themes are present in the eighteenth-century invention of rehabilitation. The rehabilitation physicians Reuben Eldar, and Miroslav Jelić write that the "true spirit of rehabilitation…started in Europe in the 18th century [24]." One of the sources for the new medical discipline of rehabilitation was orthopaedics. The French physician, Nicolas Andry (1658-1742) in setting out the principles of the new field wrote in his *Orthopédie*, translated into English in 1743, that "We are born for one another, and ought to shun having any thing about us that is shocking [25]." As such, orthopaedics was in its origins "defined primarily in terms of aesthetic improvement rather than restoration of functional ability [26]." These ideas found incorporation in new institutes and clinics, including the first orthopaedic institute, founded in Orbe Switzerland in 1780, followed by the orthopaedic hospitals in Wurzburg, Germany in 1812, Paris in 1826 and London in 1837. It would be another hundred years before the "scientific and technological transformation of orthopaedics" occurred during the twentieth century interwar period with the rise in interest in research and development [27].

The concept of physical normalcy gained further momentum in the nineteenth century with the emergence of bodily statistics [28]. It was accompanied by increasing public awareness of prosthetic devices in marketplaces. The forces behind this growth in social consciousness of artificial parts as aesthetic improvement included the use of advertising by the prosthesis industry [29]. With the rise in social awareness there were also increases in demand for functional devices, especially for industrial workers who sought to return to work after limb loss [30]. This meant the emergence of a new market for industrial workers and new meanings for their devices, versus middle class and aristocratic amputees who wished to maintain social distinction through limbs which were presentable in polite society [31]. One of the new technologies that allowed for maintenance of social distinction was a body-powered upper limb prosthesis designed by German dental surgeon Peter Baliff in 1818 [32]. It used leather straps connected to the trunk and shoulder girdle to flex fingers and extend the forearm prostheses. It offered to amputees for the first time a means to operate prosthesis with fluid body motions. It also provided the basis for subsequent improvement and adaptation throughout the nineteenth century. It was applied to above-elbow prostheses by a Dutch sculptor, Von Peterssen, in 1844. In 1867 the Comte de Beaufort redesigned it for lower cost production for use by the poor and amputee veterans of the Crimean War (1853-56). There followed numerous redesigns, including a double spring hook for holding objects, similar to that of the well-known split hook of today

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DEVELOPMENT OF A MINIATURE RATCHETING PROSTHETIC DIGIT FOR SMALL ADULTS AND CHILDREN WITH PARTIAL HAND AMPUTATION

Jacob L. Segil[1,2], Stephen Huddle[3], Richard F. *ff*. Weir[1,3]

[1]*Rocky Mountain Regional VA Medical Center*

[2]*University of Colorado Boulder*

[3]*University of Colorado | Anschutz Medical Campus*

ABSTRACT

The loss of digits from the hand is the most common upper limb amputation but there is a lack of commercially available prosthetic digits to replace what was lost. The lack of prosthetic options is especially severe for small adults of children with hand sizes that are too small for existing prosthetic componentry. Here we describe the design of a miniature ratcheting prosthetic digit for small adults and children with partial hand amputation. The design features are discussed and mechanical testing of the digits was performed. The results indicate that the miniature digit can withstand heavy-duty loads and has a failure strength of over 275 lbs while sized for a 5th percentile female hand or an 8-year old child. Soon, this miniature digit will hopefully serve a population of people with partial hand amputation that have been underserved up to this point in time.

INTRODUCTION

There are approximately 500,000 people living with minor upper limb loss in the USA **[1], [2]**. Minor upper limb loss (also partial hand amputation) is defined as the amputation of the bones distal to the wrist joint. While the field refers to these types of amputation as 'minor,' it can be a severe disability, especially if the amputation involves the thumb and/or multiple digits. In fact, partial hand amputees self-report a higher level of disability compared to other major unilateral upper limb amputees **[3]**. Furthermore, it was reported that fewer than half of partial hand amputees were able to return to the same job after amputation and most found that the prosthetic devices were insufficient to meet the demands of their work **[4]**. Amputation can cause physical, psychosocial, and economic damage to an individual and can lead to depression, anxiety, loss of self-esteem, and social isolation **[5], [6]**. While the number of individuals with partial hand amputation is 10 times more than all other categories of upper limb amputation combined, the state of available technology for this underserved patient population is relatively poor **[7]**.

Figure 1: A depiction of the miniature digits spanning lengths of 55mm (little finger) to 75mm (middle finger) on a full hand model.

Current partial hand prostheses are limited in several ways. First, they generally lack robustness, and there are frequent reports of devices breaking under normal use. Second, most current options offer a one-size-fits-all approach, which limits the acceptance by people who want a prosthesis that matches their original finger size. A complete lack of prosthetic finger options can occur for men and women with smaller anatomical size. Third, rotation of the finger about the anatomical center of rotation of the metacarpophalangeal (MCP) joint is not possible with current options resulting in a prosthesis that is frequently too long. Fourth, current prostheses require the use of the opposite hand to operate the device. There is a need for a durable, scalable, single-handedly operable prosthetic finger which is anatomically suitable for small adults or children.

The lack of prosthetic limb systems for small adults and children with upper limb amputation is a problem within our field. Even data concerning the prevalence of upper limb loss among women in particular is lacking **[7]**. Ziegler et al. estimated limb loss in the United States in 2008 and determined that women comprised 35% of all persons with upper and lower limb loss **[1]**. More specifically, Atkins et al performed a thorough survey of persons with upper-limb loss in 1996 and found that women comprised 22% of persons with upper limb loss **[8]**. Based on this percentage (22% women) while the majority of people with upper limb amputation are male, there is still a large population of women, approximately 110,000 women **[1], [2]** who are in need of better prosthetic options at the finger level.

The differences between male and female hand anatomy are great; a 50th percentile female middle finger length (101mm) is equivalent to the 1st percentile male middle finger length (102mm) **[9]**. Also, the 1st percentile male middle finger length (102mm) is 13mm longer than the 1st percentile female middle finger length (89mm). These differences essentially preclude the use of most current partial hand prosthetic devices for women who are not in the 5th percentile or greater. With this effort, we sought to design a miniature prosthetic finger with 55mm length that would provide a valuable device for 5th percentile women and 5-year-old children (Figure 1) **[10]**.

DESIGN METHODS

The design of the miniature digit was based upon our prior work on the *Point Digit* now commercialized by Point Designs LLC (Lafayette, CO). The miniaturization of the digit affects the function of the mechanical systems within it. Multiple mechanisms within the current design were scaled with the size of the finger including 1) the ratchetting mechanism, 2) the bump release mechanism, and 3) the push-button release mechanism. The tolerances involved were redefined for the miniature mechanisms to provide the same amount of reliability and function using a repeatable manufacturing method. The bump release unlocks the ratchetting mechanism allowing the spring to bring the finger to full extension. The “props” within the bump release mechanism are used to unlock the ratchetting mechanism bringing the finger to full extension.

The current, full-sized Point Digit weight specification is no more than 70 g (1/8 lb.) based on an estimate of anatomical weight where the hand makes up 0.75% of body weight [11]. In the case of a 175 lbs. male whose hand would weigh about 1.25 lb., and assuming the palm makes up 50% of that weight, each finger and thumb would be 1/8 lb. (~66 g). The Point Digit prototype weighs between 45 g (80 mm length) to 55 g (120 mm length), satisfying the weight requirement for male hands. We sought to reduce the weight of the Point Digit by 50% to a weight of 25g in order to fit women with body weight as little as 90 lbs. To achieve this, we will print the finger in titanium as well as light-weighting the componentry using direct metal laser sintering techniques.

EXPERIMENTAL METHODS

We evaluated each finger on a bench-top material-testing machines (MTS) and custom finger cyclers to establish if the design meets specifications. To create a clinically-ready prosthetic finger, we need the miniature digit to be durable, robust, and able to withstand high forces. The mechanical testing spanned static loading conditions, dynamic loading conditions, and dynamic unloaded conditions. Each test was conducted with loads applied to the palmar surface of the distal phalange, the palmar surface of the medial phalange, and lateral surface of the distal phalange. The static loading test applied loads of 66N on the palmar and lateral surface of the distal phalange as this is the minimum a prosthetic hand should be able to generate [12] and loads of 98N on the palmar surface of the medial phalange as this is approximately the weight of a bag of groceries (10kg). After achieving this minimum specification, each finger was tested to failure with loads applied to each finger surface. The dynamic loading test applied a repetitive fatigue load of the same magnitude of the static loading testing at a rate of 1Hz for 10,000 cycles. The dynamic unloaded test cycled the digit through 250,000 flexion and extension cycles. This simulated the unloaded use of the digit over 3 years of use assuming ~30 grip changes per hour, 8 hours of wear time per day, and 365 days of use per year. A custom made cycle testing machine used an actuator to flex the digit into full-flexion and then relied on the spring-back mechanism within the digit to fully extend the digit. This test ensured the mechanism can withstand long-term use while maintaining the ratcheting and spring-back functions. The successful completion of these specifications will match the mechanical specifications of the Point Digit.

RESULTS

A battery of verification tests was performed to ensure that all design specifications of the miniature prosthetic digit had been met. In all tests, the digit met or exceeded the specification (Table 1). In some cases, the digit exceeded the specification by several factors. The miniature digit met the mass and length requirements of a 5th percentile female as well as a 5-year old child. In the most demanding test, the static load test at the distal tip of the finger, the static strength of the Point Digit at the distal fingertip exceeded the specification by 18 times (1,242 N compared to 66 N, 279 lb compared to 15 lb). The dynamic load tests were all successful in that the digits did not show any visible signs of wear, damage, or loss of function. The dynamic cycle test was not completed at the time of this publication, but has proceeded without adverse events at this time. These results indicate that the miniaturization of the digit did not cause a decrease in mechanical performance of the mechanisms involved in the ratcheting prosthetic digit.

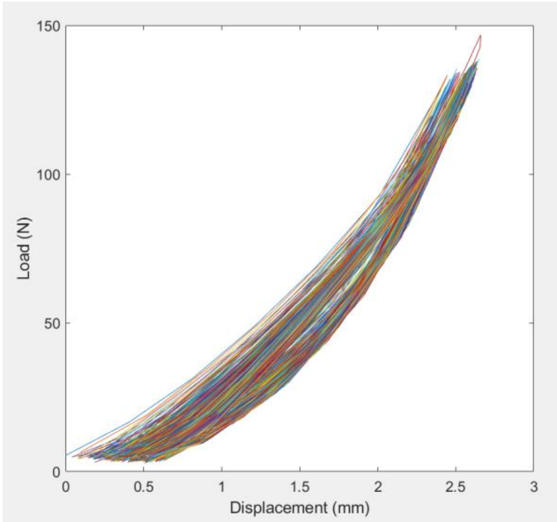

Figure 2: Exemplary data from the dynamic load test applied to the medial phalange. 500 loading cycles are shown out of the 10,000 cycles performed.

Table 1. Mechanical Testing Results

	Mass	**Length**	**Dynamic Cycle Test**	**Dynamic Load Testing (10,000 cycles @ 1Hz)**			**Static Load Testing (Load to failure)**		
Specification	25g – 35g	55mm – 75mm	250,000 cycles	Distal 66 N	Medial 98 N	Lateral 66 N	Distal ≥66 N	Medial ≥98 N	Lateral ≥66 N
Testing Result	30g	55mm – 75mm	N/A	10,000 cycles at 66 N	10,000 cycles at 98 N	10,000 cycles at 98 N	1,242 N	≥200 N	≥200 N
Meets Specification	✓	✓	--	✓	✓	✓	✓+++	✓+	✓+
Exceeds Specification by	--	--	--	1x	1x	1x	**18.8x**	2x	2x

DISCUSSION

The mechanical design of miniature prosthetic fingers for small adults and children with partial hand amputation requires a great reduction in size of the internal mechanisms involved in the devices. This work details the creation of a robust device which can withstand heavy-duty use for these under-served patient populations. A battery of mechanical tests confirmed that the miniature digit could withstand clinically-relevant loads and cycles of use. The success of this mechanical design was accomplished using rapid-manufacturing technology like direct-metal laser sintering 3D printing. This manufacturing method enables complex componentry to be created with internal geometries and in-situ functions that are not possible with other manufacturing techniques. Furthermore, the 3D printing technology allows for the digit to be easily scaled between 55mm – 75mm and thereby produce appropriate lengths for a variety of patients. Future work will produce miniature digits for small adults in a laboratory testing session as well as a take-home clinical trial. Then, relevant outcome measures will be produced in order to detail the utility of the miniature digit during activities of daily living.

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