Continuous Proportional Myoelectric Control of an Experimental Powered Lower Limb Prosthesis During Walking Using Residual Muscles

by

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ABSTRACT

Current robotic lower limb prostheses rely on intrinsic sensing and finite state machines to control ankle mechanics during walking. State-based controllers are suitable for stereotypical cyclic locomotor tasks (e.g. walking on level ground) where joint mechanics are well defined at specific gait phases (i.e. states) and state transitions are easily detected. However, state-based controllers are not ideal for non-stereotypical acyclic tasks (e.g. freestyle dancing) where joint mechanics cannot be predefined and transitions are unpredictable. An alternative to state-based control is to utilize the amputee's nervous system for myoelectric control. A robotic lower limb prosthesis that uses continuous proportional myoelectric control would allow the amputee to adapt their ankle mechanics freely. One potential source for myoelectric control is the amputee's residual muscles. I conducted four studies to examine the feasibility of using residual muscles for continuous myoelectric control during walking.

In my first study, I demonstrated that it is possible to record residual electromyography from amputees during walking that are viable for continuous myoelectric control. My results showed that the stride-to-stride variability of residual and intact muscle activation patterns was similar. However, residual muscle activation patterns were significantly different across amputee subjects

and significantly different than corresponding muscles in intact subjects. In my second study, I built and tested an experimental powered transtibial prosthesis and demonstrated that an amputee subject was able to walk using continuous proportional myoelectric control to alter prosthetic ankle mechanics. In my third study, I showed that five amputee subjects were able to adapt their residual muscles to walk using continuous proportional myoelectric control. With visual feedback of their control signal, amputees were able to generate higher peak ankle power walking with the experimental powered prosthesis compared to their prescribed prosthesis. In my fourth study, I conducted a user experience study and found that despite challenges with the device user interface, walking with continuous proportional myoelectric control gave amputees a sense of empowerment and embodiment. The results of my studies demonstrated the advantages and disadvantages of using continuous proportional myoelectric control for a powered transtibial prosthesis and suggest how next generation prostheses can build upon these findings.

Chapter 1: Introduction

There are a growing number of amputees in the United States and around the world who depend on prosthetic devices and prosthetic technology to regain mobility and thereby improve their quality of life. In 2005, there were an estimated 1.5 million amputees in the United States of which 65 percent were lower limb amputees [1]. Lower limb amputees suffer from limited mobility largely as a result of localized pain in the residual limb and walking fatigue [2]. During walking, unilateral transtibial amputees expend 20-30% more energy compared to able-bodied persons walking at the same speed [3] and unilateral transfemoral amputees experience an additional 25% increase in energy expenditure [4]. For this reason, a long-standing challenge for new lower limb prostheses is to demonstrate significant reductions in metabolic energy expenditure. Energy-storing-and-returning prosthetic feet are designed to store energy during stance and return energy during late-stance to assist with forward propulsion. Despite recent advances, none of the currently available energy-storing-and-returning prosthetic feet have been shown to significantly decrease the metabolic cost of walking in transtibial amputees compared to conventional feet [5]. The development of more intelligent controllers that automatically alter the mechanical output of a powered prosthesis according to gait phase has implications of reducing metabolic cost during locomotion. For this reason, the lower limb prosthetic research focus has shifted towards to the development of powered prosthetic feet to mimic human ankle behavior.

Current robotic lower limb prosthesis controllers use finite state machines and intrinsic sensing to alter prosthetic mechanics during walking [6-13]. Force and motion sensors provide information about the prosthesis to a control algorithm that then shifts actuators into predefined states for facilitating locomotion. These state-based controllers using intrinsic sensing are typically tuned to best perform level-ground walking at a selected speed. While state-based controllers can be relatively successful for level-ground walking, they perform less than ideally for a wider range of locomotor tasks (i.e. acyclic, non-stereotypical tasks) and for discrete tasks like a sit-to-stand motion. Powered lower limb prostheses are likely to have improved functionality and effectiveness if alternative control methods can provide a more adaptive command of the prosthesis mechanics.

Myoelectric control has been available for upper-limb prostheses for many years, but it usually places a relatively high cognitive load on the user [14]. One explanation for the relatively high cognitive load of using myoelectric upper limb prostheses is that the muscle activation patterns used to decipher movement intent are not naturally occurring (i.e. the amputee user needs to learn muscle activation patterns that are not intuitive) [15, 16]. An alternative is to exploit naturally occurring muscle activation signals as a control source for powered prostheses. This naturally occurring muscle activity could provide a more physiological and adaptable interface for amputees to control powered prostheses.

Muscle activation signals recorded from residual limb muscles could provide an intuitive command source for myoelectric control of lower limb powered prostheses. One advantage of using residual limb muscles is that they are already embedded into physiological pathways and motor synergies relevant to joint biomechanics. Despite ongoing developments in myoelectric control, there are no commercially available myoelectric controllers for powered lower limb prostheses [5, 17, 18]. Recently, a few research groups have begun to use muscle activity signals from the residual lower limb muscles for input to powered prostheses [19-24]. Myoelectric controllers for lower limb powered prostheses can aid in identifying and transitioning between phases of the gait cycle and different locomotor tasks [23]. Herr and colleagues have also been able to integrate myoelectric signals into an intrinsic sensing state controller to discretely scale the relative strength of actuator responses within specific states [24]. None of these approaches, however, provide the user with continuous control of joint dynamics that approaches typical human physiology. To the best of my knowledge, no research groups have created and tested a continuous myoelectric controller using residual muscle activity to directly control a powered lower limb prosthesis.

The goal of my dissertation was to determine the feasibility of using residual limb muscle activity from lower limb amputees during walking for continuous proportional myoelectric control of a transtibial powered prosthesis. Chapters 2-5 contain four complete manuscripts that can be read as individual studies. In my first study (Chapter 2), I demonstrated that it is feasible to record robust and reliable residual muscle activation signals at the limb-socket interface across a wide range of walking speeds. In my second study (Chapter 3), I designed, built, and tested an experimental powered prosthesis that uses continuous proportional myoelectric control from residual limb muscles to directly alter ankle mechanics during walking. In my third study (Chapter 4), I examined how amputee subjects adapt their residual muscle activation patterns and quantified their locomotor adaptation when walking with the experimental powered prosthesis using continuous proportional myoelectric control. In my fourth study (Chapter 5), I conducted a user experience study to better understand how the amputee subjects interacted with the experimental powered prosthesis both physically and mentally.

Chapter 2: Muscle Activation Patterns During Walking from Transtibial Amputees Recorded within the Residual Limb-Prosthetic Interface

This chapter has been previously published:
S. Huang and D. P. Ferris, "Muscle activation patterns during walking from transtibial amputees recorded within the residual limb-prosthetic interface," *Journal of Neuroengineering and Rehabilitation,* 9, 55, 2012.

ABSTRACT

Powered lower limb prostheses could be more functional if they had access to feedforward control signals from the user's nervous system. Myoelectric signals are one potential control source. The purpose of this study was to determine if muscle activation signals could be recorded from residual lower limb muscles within the prosthetic socket-limb interface during walking. We recorded surface electromyography from three lower leg muscles (*tibilias anterior*, *gastrocnemius medial head*, *gastrocnemius lateral head*) and four upper leg muscles (*vastus lateralis*, *rectus femoris*, *biceps femoris*, and *gluteus medius*) of 12 unilateral transtibial amputee subjects and 12 non-amputee subjects during treadmill walking at 0.7, 1.0, 1.3, and 1.6 m/s. Muscle signals were recorded from the amputated leg of amputee subjects and the right leg of control subjects. For amputee subjects, lower leg muscle signals were recorded from within the limb-socket interface and from muscles above the knee. We quantified differences in the muscle activation profile between amputee and control groups during treadmill walking using cross-correlation analyses. We also assessed the step-to-step inter-subject variability of these profiles by calculating variance-to-signal ratios. We found that amputee subjects demonstrated reliable muscle recruitment signals from residual lower leg muscles recorded within the prosthetic socket during walking, which were locked to particular phases of the gait cycle. However, muscle activation profile variability was higher for amputee subjects than for control subjects. Our findings suggest that robotic lower limb prostheses could use myoelectric signals recorded from surface electrodes within the socket-limb interface to derive feedforward commands from the amputee's nervous system.

INTRODUCTION

Recent advances in robotic technology have allowed for the development of powered lower limb prostheses that improve ambulation for amputees. A major feature of these new devices is the ability to interject mechanical power into the gait cycle to replace the mechanical power that is lost due to missing biological muscles. Hugh Herr's research group at the Massachusetts's Institute of Technology has developed a robotic ankle that uses a finite state controller to modulate ankle dynamics during gait and add power to the trailing limb during push off [6, 7, 25]. The prosthesis uses intrinsic sensing of kinetics and kinematics (e.g., heel- and toe-contact, ankle angle, and ankle torque) to determine when to transition between gait phases during walking. Their powered prosthesis resulted in lower metabolic cost compared to traditional passive elastic prostheses for level ground walking [9]. In addition to a robotic ankle, they have developed a variable impedance robotic knee that uses intrinsic sensing and a finite state controller to modulate knee stiffness during level ground walking [26]. Michael Goldfarb's research group at Vanderbilt University has developed a robotic knee and ankle for transfemoral amputees that also uses intrinsic sensing and finite state control [11-13]. Tom Sugar's research group at Arizona State University developed a powered ankle that relies on elastic elements to store energy and amplify mechanical power generated by the actuator [10]. It uses intrinsic sensing to detect heel strike and then the controller initiates a predetermined gait pattern. This sampling of robotic prostheses is representative of the intrinsic sensing approaches that are beginning to be utilized for prosthetic control [5, 18].

There are advantages and disadvantages of controlling prosthetic lower limbs via intrinsic sensing. An advantage of prosthetics that rely on kinetic and kinematic sensing to infer user intent is that all of the sensors and associated computational hardware are built directly into the prosthetic. The interface with the human is purely mechanical, which simplifies socket design. These prosthetics generally have low step-to-step variability due to the robustness of the finite state controllers and the low sensor noise. Controllers based on intrinsic sensing tend to work well for stereotyped or cyclical tasks, such as gait. One of the inherent drawbacks of these devices is that control based on intrinsic sensing is not very good at aperiodic or highly variable motor tasks. For example, going up on the toes to reach a higher shelf would be very difficult for a state-based controller to perform using intrinsic sensing. Similarly, tasks with highly variable step-to-step kinematics such as traversing obstacles in the terrain, traversing unstable terrain, or negotiating through a crowd of people, or dealing with a variety of natural surfaces like sand and rocks would be difficult to deal with using intrinsic sensing alone.

An alternative to controllers that rely solely on intrinsic kinematic and kinetic sensing is to directly connect the prosthesis dynamics to the user's nervous system via electromyography [14, 16, 27]. Myoelectric control has been implemented for powered upper limb prostheses. High costs have limited widespread acceptance of these devices but cost will continue to fall with continued technological advances. A more lasting obstacle to widespread acceptance of powered upper limb prostheses is the degrees of freedom that must be controlled. The human hand and wrist have more than 20 mechanical degrees of freedom but upper limb prostheses usually rely on fewer than 6 myoelectric control sources. This limits the ability for users to accurately and reliably control prosthesis mechanics. For the lower limb, fewer mechanical degrees of freedom are necessary to provide functional motor ability. For a transtibial amputee, active mechanical plantar flexion/dorsiflexion and passive foot elasticity can provide a huge energetic improvement compared to passive lower limb prostheses [9].

Controlling a limited number of mechanical degrees of freedom with myoelectric signals is feasible. Transfemoral amputees can learn to volitionally control virtual knee/ankle joint movements using myoelectric control signals from residual thigh muscles while seated and not wearing their prosthesis [19, 21]. In addition, transtibial amputees can learn to volitionally activate residual muscles during the swing phase of walking to switch between level-ground walking and stair-descent locomotion modes [6]. To the best of our knowledge, this is the only case where myoelectric signals have been recorded from within the socket-limb interface during walking and used for user movement intent recognition.

To implement more robust myoelectric controllers for transtibial prostheses, it is important to assess lower leg electromyographic signal quality, variability, and adaptability during amputee gait. In the near future, it may be possible to use intramuscular electromyography sensors (IMES) to transmit electromyographic signals through the socket interface without breaking the skin [28-30]. These IMES would make it feasible to implement a wide range of myoelectric control methods with powered prostheses. However, rather than waiting for these IMES to be approved for human testing, we have recorded electromyography from lower leg muscles of transtibial amputees within the socket interface using surface electrodes. The purposes of this study were 1) to determine if surface electromyography signals can be recorded from residual lower leg muscles inside the prosthetic socket during walking, and 2) to quantify differences in muscle activation patterns between amputee and non-amputee subjects during walking.

METHODS

Subjects

We recruited twelve unilateral transtibial amputee subjects (10 male, 2 female; age=46±18 yrs.; height=175±8 cm.; mass=81±10 kg.; mean±s.d.) and twelve non-amputee subjects (8 male, 4 female; age=37±15 yrs.; height=173±15 cm.; mass=76±18 kg.) to participate in this study. All subjects were free of musculoskeletal and cardiovascular conditions that would limit their ability to walk safely on a treadmill. All amputee subjects had been using their prosthesis for at least six months, were accustomed to walking on their prosthesis all day, and could walk comfortably without the use of an additional ambulatory aid. Amputee subject details are listed in Table 2-1.

Instrumentation

We collected surface electromyography (EMG) from seven lower limb muscles: *tibialis anterior*, *gastrocnemius medial head*, *gastrocnemius lateral head*, *vastus lateralis*, *rectus femoris*, *biceps femoris*, and *gluteus medius*. We recorded EMG signals at 1000 Hz using pre-amplifier electrodes (Biometrics Ltd, SX230) from

Table 2-1. Amputee Subject Details

Subject	Reason for Amputation	Age (yrs.)	Post-Amputation (yrs.)	Fastest Walking Trial (m/s)
A01	Cancer	20	11	1.6
A02	Trauma	49	7	1.6
A03	Cancer	18	6	1.6
A04	Trauma	66	7	1.0
A05	Trauma	31	1	1.3
A06	Trauma	55	1	1.0
A07	Trauma	56	40	1.6
A08	Trauma	44	5	1.0
A09	Diabetes	65	10	1.0
A10	Trauma	61	41	1.6
A11	Trauma	59	8	1.6
A12	Trauma	27	3	1.6

the amputated leg of amputee subjects and the right leg of non-amputee subjects. For upper leg muscles of all subjects and lower leg muscles of control subjects, we placed the electrode over the muscle belly and along the direction of the muscle fibers. To determine the location and orientation of each electrode, we palpated each muscle area while subjects performed a series of voluntary muscle activations. For the lower leg muscles (*tibialis anterior*, *gastrocnemii*) of amputee subjects, we marked a grid of potential recording sites on the skin surface over each muscle that we identified by palpating underlying tissue and bone. We avoided sensitive skin areas and bony protuberances. We subjectively ranked each recording site on the grid based on muscle quality (perceived by palpating the muscle area during voluntary muscle activations). We positioned one electrode over the "best" recording site on each muscle and subjects donned their prosthesis and walked around the laboratory to assess comfort. We did not make any modifications to their prosthesis. To adjust socket fit, subjects changed the thickness of socks they wore between the gel liner and prosthesis socket. If subjects expressed discomfort with an electrode, we shifted the position slightly or chose a secondary recording site. Once the recording sites were finalized, we placed silicone putty around the edges of the electrodes and secured the electrodes to the skin using Tegaderm™ dressing. The silicon putty minimized skin irritation around the electrode edges. The sensor placement procedure is outlined in Figure 2-1. We placed the ground electrode on the *lateral malleolus* of the intact leg for amputee subjects and the *lateral malleolus* of the right leg for non-amputee subjects.

We recorded ground reaction forces in the vertical, medial-lateral, and fore-aft directions at 1000 Hz using a custom-built instrumented split-belt treadmill [31]. We defined heel-strike and toe-off events from vertical ground reaction force.

Protocol

The first part of the test protocol assessed the subject's ability to differentiate plantar flexor and dorsiflexor muscle activation. Subjects performed maximum voluntary activation trials where they tried to isolate the activation of their *tibialis anterior* (dorsiflexion trial) and *gastrocnemii* (plantar flexion trial) muscles. Subjects were seated upright on a raised platform so that their feet did not contact the ground during the maximum voluntary activation trials. To obtain maximal activation of the *tibialis anterior*, we instructed subjects to point their feet and toes towards the ceiling as hard as possible and sustain muscle

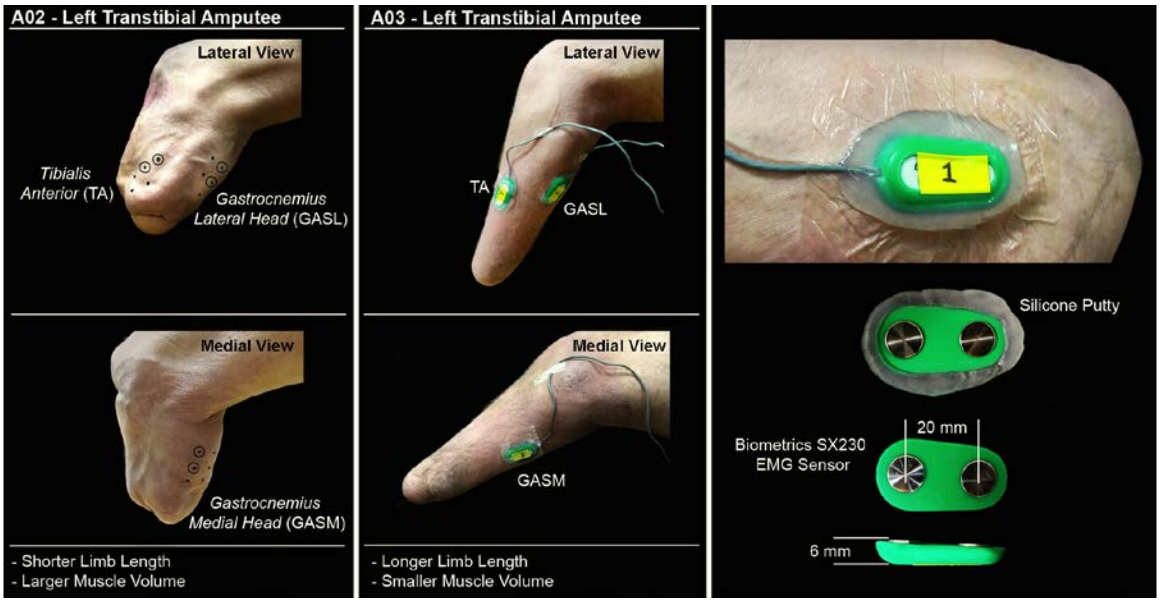

Figure 2-1. Surface Electrode Placement for Residual Lower Leg Muscles.
Tibialis Anterior (TA), *Gastrocnemius Medial Head* (GASM), *Gastrocnemius Lateral Head* (GASL). Two amputee subjects (*A02*, *A03*) show the extent of variation in lower leg shape of our amputee subjects. Subject *A02* (49 year old, amputation due to trauma at age 42) has a relatively short lower leg with relatively large muscle volume. In comparison, subject *A03* (18 year old, amputation due to cancer at age 12) has a longer lower leg with smaller muscle volume. As shown on subject *A02*, a grid of potential electrode locations was marked on the skin surface over the lower leg TA, GASM, and GASL. From each grid, the primary electrode site was determined by palpation during voluntary contractions of the muscle. Electrodes were placed over the primary electrode site and the gel liner and socket were worn over the electrodes. No modifications to the gel liner or socket were made. Socks of varying thickness were used to adjust socket-fit. Subjects were asked to walk around the laboratory to assess comfort at the primary electrode sites. If there was discomfort, electrodes were repositioned slightly or secondary sites were selected. The final electrode sites for subject *A02* are circled. After the electrode sites were finalized, silicone putty was placed around the electrode and the electrode was secured to the skin using a piece of Tegaderm™ dressing.

activation at maximum dorsiflexion. To obtain maximal activation of the *gastrocnemii*, we instructed subjects to point their feet and toes towards the ground as hard as possible and sustain muscle activation at maximum plantar flexion. All ankle movements were performed bilaterally. We instructed amputee subjects to activate their lower leg muscles as if they had an intact ankle and foot. During practice trials, we displayed real time EMG signals to amputee subjects to provide feedback on the level of muscle activation. Once EMG signals appeared consistent, we recorded three repetitions for each maximum voluntary activation task. For each repetition, we asked the subjects to sustain the maximum voluntary activation for five seconds then rest with muscles fully relaxed for five seconds.

The second part of the test protocol assessed muscle activation patterns during walking. Subjects walked on a treadmill at four speeds (0.7, 1.0, 1.3, and 1.6 m/s) for two minutes at each speed. Not all subjects were able to walk at the two faster speeds. To determine the fastest walking trial that subjects could complete safely, we asked each subject to practice walking on the treadmill starting at the slowest speed. If they could walk comfortably at the given speed, we increased the treadmill speed gradually to the next level. We continued this until the fastest treadmill speed was reached or until the subject could no longer maintain walking speed. All subjects completed the 0.7 and 1.0 m/s trials. Eight of the twelve amputee subjects and eleven of the twelve control subjects completed the 1.3 m/s trial. Seven of the twelve amputee subjects and eleven of the twelve control subjects completed the 1.6 m/s trial.

Signal Processing

We performed all signal processing and statistical analyses using the R computing environment (R Development Core Team, 1999). We processed EMG signals using two separate methods. To look at raw EMG, we applied a high-pass filter (bidirectional Butterworth, 4^{th} order, 50 Hz cutoff frequency) and then demeaned the signal. We chose a cutoff frequency of 50 Hz to ensure that motion artifacts were attenuated. To analyze the frequency content of the signal, we calculated a smoothed periodogram estimated by a discrete Fourier transform and filtered using Daniell smoothers (single span of length 5). We calculated an empirical cumulative distribution function of the power spectrum to compare the distribution of frequency content between the amputee and control groups.

For the maximum voluntary activation trials, we performed frequency analysis of the *tibilias anterior* and *gastrocnemii* EMG for two seconds of sustained activation. For each subject, we selected the repetition where the maximum amplitude of the rectified signal (high-pass filtered and demeaned) was the greatest across trials. For some amputee subjects, the residual limb *tibialis anterior* was activated more than the *gastrocnemii* during the plantar flexion trial and vice versa during the dorsiflexion trial. For 1.0 m/s walking, we performed frequency analysis of the *tibilias anterior* and *gastrocnemii* for a single gait cycle. For each subject, we selected the gait cycle where the variance of the signal (high-pass filtered and demeaned) was closest to the mean variance of all cycles.

To quantify muscle activation profiles, we calculated EMG intensity using a wavelet decomposition method [32]. We calculated an intensity curve by summing across wavelets 4 (center frequency=62.1 Hz) through 11 (center frequency=395.5 Hz) in time. This method was chosen over other methods (e.g. generating a linear envelope using a low-pass filter) because the intensity curve provided a more distinct profile, specifically at transitions between baseline and activation. We divided the intensity curve into cycles defined by consecutive heel strike events. We normalized time by interpolating over 500 equally spaced points per cycle using cubic splines, and we normalized the amplitude to the maximum amplitude across all walking speeds. We calculated a mean intensity curve from 40 consecutive time- and amplitude-normalized cycles. To quantify the repeatability of the recorded EMG signals, we calculated a variance-to-signal ratio (VSR) as the sum of the signal variance over the sum of the signal mean squared across the 40 consecutive normalized intensity curves: $VSR = \frac{\sum_{i=1}^{500} \sigma_i^2}{\sum_{i=1}^{500} \mu_i^2}$ [33]. To quantify differences in EMG shape, we used mean intensity curves to calculate normalized cross-correlations with zero time lag [34] between: 1) control group grand mean and control subject mean $(\rho_{\bar{X}X_i})$, 2) control group grand mean and amputee subject mean $(\rho_{\bar{X}Y_i})$, and 3) amputee group grand mean and amputee subject mean $(\rho_{\bar{Y}Y_i})$. For cross-correlations $(\rho_{\bar{X}X_i})$ and $(\rho_{\bar{Y}Y_i})$, individual subject data was excluded from the group mean.

Normalized cross-correlations were calculated for EMG from all seven muscles using the subset of subjects who completed all four walking speeds.

Statistical Analyses

We performed two separate ANOVAs to determine if there were significant differences in median EMG frequency between subject groups during either maximum voluntary activations or treadmill walking at 1.0 m/s. (model: *median frequency~muscle+group*). We performed another ANOVA to determine if there were significant differences in median EMG frequency between maximum voluntary activation and treadmill walking (factor: *task*) at 1.0 m/s for lower leg muscles only (model: *median frequency~muscle+group*task*). We performed two ANOVAs to determine if there were significant differences in cross-correlation (*R*-value) between subject groups (model: *R*-value~*muscle+group*). For the first ANOVA, the independent variable was $(\rho_{\bar{X}X_i})$ for control subjects and $(\rho_{\bar{X}Y_i})$ for amputee subjects. For the second ANOVA, the independent variable was $(\rho_{\bar{X}X_i})$ for control subjects and $(\rho_{\bar{Y}Y_i})$ for amputee subjects. For all ANOVAs, if factors of interest were significant (p<0.05), we performed a Tukey's Honestly Significant Difference test to determine which contrasts were significant (p<0.05).

RESULTS

Maximum Voluntary Activation of Lower Leg Muscles

Amputee subjects were able to volitionally activate their lower leg muscles during the maximum voluntary activation trials but the relative activation of agonist and antagonist muscles was not consistent across subjects (Figure 2-2A). All control subjects had high and well-sustained agonist muscle activation and low antagonist muscle activation during the trials. Five amputee subjects had muscle activation patterns similar to controls (e.g., Figure 2-2A, subjects *A05*, *A06*, *A07*, *A09*, and *A10*). These subjects had a range of 1–41 years since amputation (Table 2-1). Two amputee subjects had high activation of both agonist and antagonist muscles during plantar flexion and little to no activation of agonist or antagonist muscles during dorsiflexion (e.g., Figure 2-2A, subjects *A02* and *A08*). Although most amputee subjects were able to sustain activation levels as well as control subjects, some had difficulty maintaining activation levels (e.g., Figure 2-2A, subjects *A01* and *A04*).

Lower Leg EMG During Walking

During treadmill walking, *tibialis anterior*, *gastrocnemius medial head*, and *gastrocnemius lateral head* activation patterns in amputee subjects had much higher inter-subject variability and were substantially different than the patterns of the control subjects (Figure 2-3A, Figure 2-4A, Figure 2-5). The high inter-subject variability in amputee EMG patterns is demonstrated by a significant difference (ANOVA, p<0.001) in EMG pattern cross-correlation between the amputee individual data vs. amputee mean, compared to the control individual data vs. the control mean $(\rho_{\bar{Y}Y_i}, \rho_{\bar{X}X_i})$ (Table 2-2). Mean cross-correlations for individual amputee EMG patterns vs. the amputee mean $(\rho_{\bar{Y}Y_i})$ ranged from 0.20-0.53 for the *tibialis anterior*, *gastrocnemius medial head*, and *gastrocnemius lateral head* (Table 2-2). In comparison, mean cross-correlations for individual control EMG patterns vs. the control mean $(\rho_{\bar{X}X_i})$ ranged from 0.73-0.92 for the same muscles (Table 2-2). In addition to the difference in inter-subject variability, the cross-correlations also provide evidence of the difference in shape of the EMG activation patterns between amputee and control subjects. There was a significant difference (ANOVA, p<0.001) in EMG pattern cross-correlation between the amputee individual data vs. control mean, compared to the control individual data vs. control mean $(\rho_{\bar{X}Y_i}, \rho_{\bar{X}X_i})$ (Table 2-2). In the amputee group, mean cross-correlations against the control mean $(\rho_{\bar{X}Y_i})$ ranged from –0.33 to 0.48 for the *tibialis anterior*, *gastrocnemius medial head*, and *gastrocnemius lateral head*. In the control group, mean cross-correlation against the control mean $(\rho_{\bar{X}X_i})$ ranged from 0.73-0.92 for the same muscles.

Upper Leg EMG during walking

Compared to lower leg muscles, upper leg muscle activation patterns during walking were more similar between amputee and control subjects (Figure 2-6A, Figure 2-7A, Figure 2-8). There was no significant difference in inter-subject variability between amputees and controls for the *vastus lateralis* and *rectus femoris* $(\rho_{\bar{Y}Y_i}, \rho_{\bar{X}X_i})$; post-hoc *t*-test (p>0.05) (Table 2-2). Mean cross-correlation for individual amputee EMG patterns vs. the amputee mean $(\rho_{\bar{Y}Y_i})$ for these muscles ranged from 0.66-0.90 (Table 2-2). In comparison, mean cross-correlation for individual control EMG patterns vs. the control mean $(\rho_{\bar{X}X_i})$ ranged from 0.63-0.90 for the same muscles (Table 2-2). For the *biceps femoris* and *gluteus medius*, there was a significant difference (post- hoc *t*-test p<0.001) in EMG pattern cross-correlation between the amputee individual data vs. amputee mean, compared to the control individual data vs. the control mean $(\rho_{\bar{Y}Y_i}, \rho_{\bar{X}X_i})$.

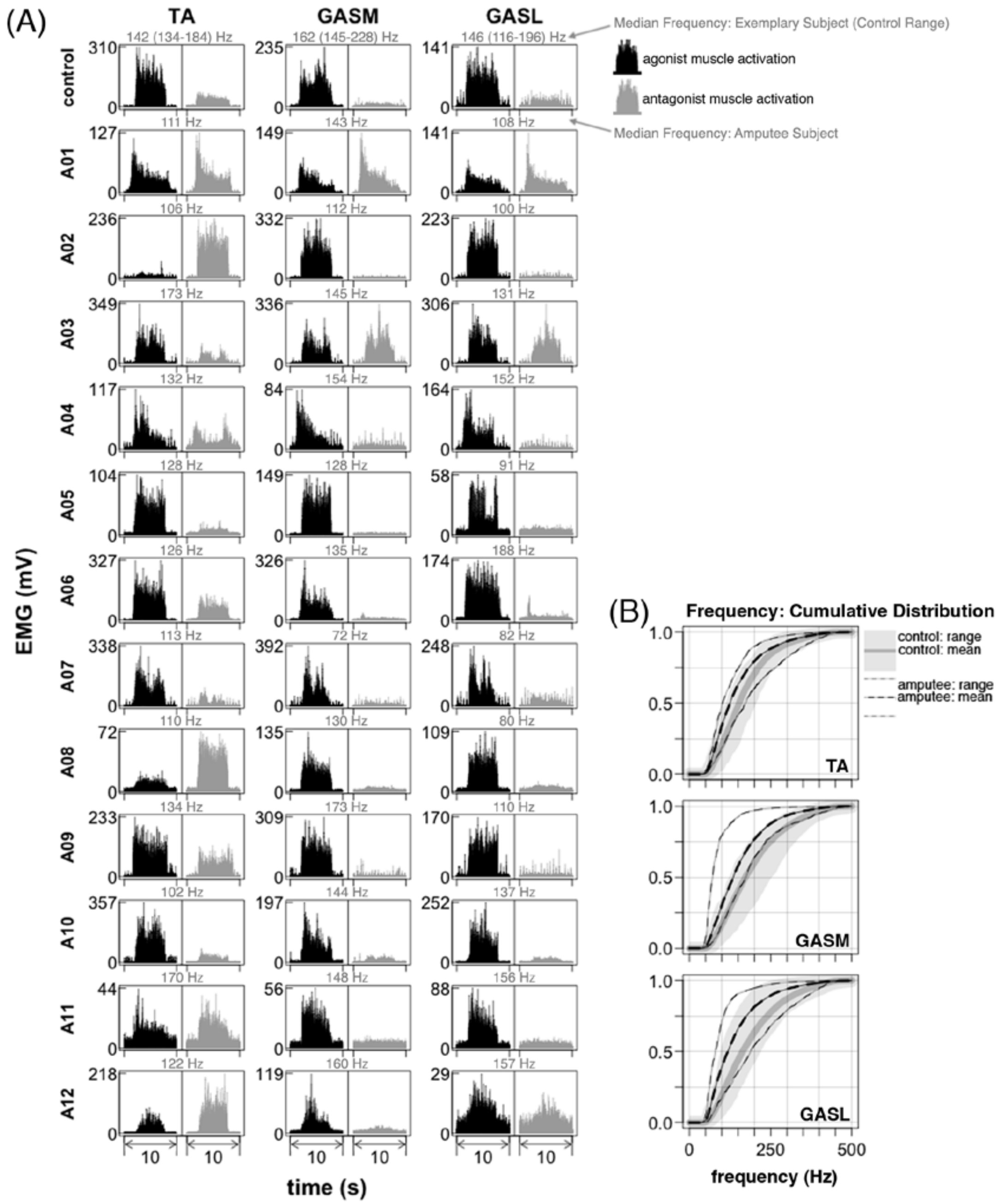

Figure 2-2. Lower Leg EMG Maximum Voluntary Activation.
Tibialis Anterior (TA), *Gastrocnemius Medial Head* (GASM), *Gastrocnemius Lateral Head* (GASL). (A) EMG during maximum voluntary activation of TA, GASM, and GASL muscles during seated dorsiflexion and plantar flexion. Data shown for one exemplary control subject and twelve amputee subjects. Signals are high-pass filtered, demeaned, and rectified. Signals in black indicate that the muscle is agonist to ankle movement. Signals in gray indicate that the muscle is antagonist to ankle movement. Median frequency during maximum voluntary activation (agonist or antagonist depending on which activation had the greatest amplitude) is shown in gray. In control subjects, there was high agonist muscle activation (black) and low antagonist muscle activation (gray). This activation pattern was not consistent in amputee subjects. Amputee subjects *A02* and *A08* had little to no lower leg muscle activation during dorsiflexion and high activation of TA, GASM, and GASL muscles during plantar flexion. *A01* had activation of all lower leg muscles for both dorsiflexion and plantar flexion, but activation was not well sustained. Some amputee subjects had activation patterns similar to controls (*A05*, *A06*, *A07*, *A09*, *A10*). (B) Empirical cumulative density function of EMG power spectrum. Lines shown for group means, boundaries indicate group range.

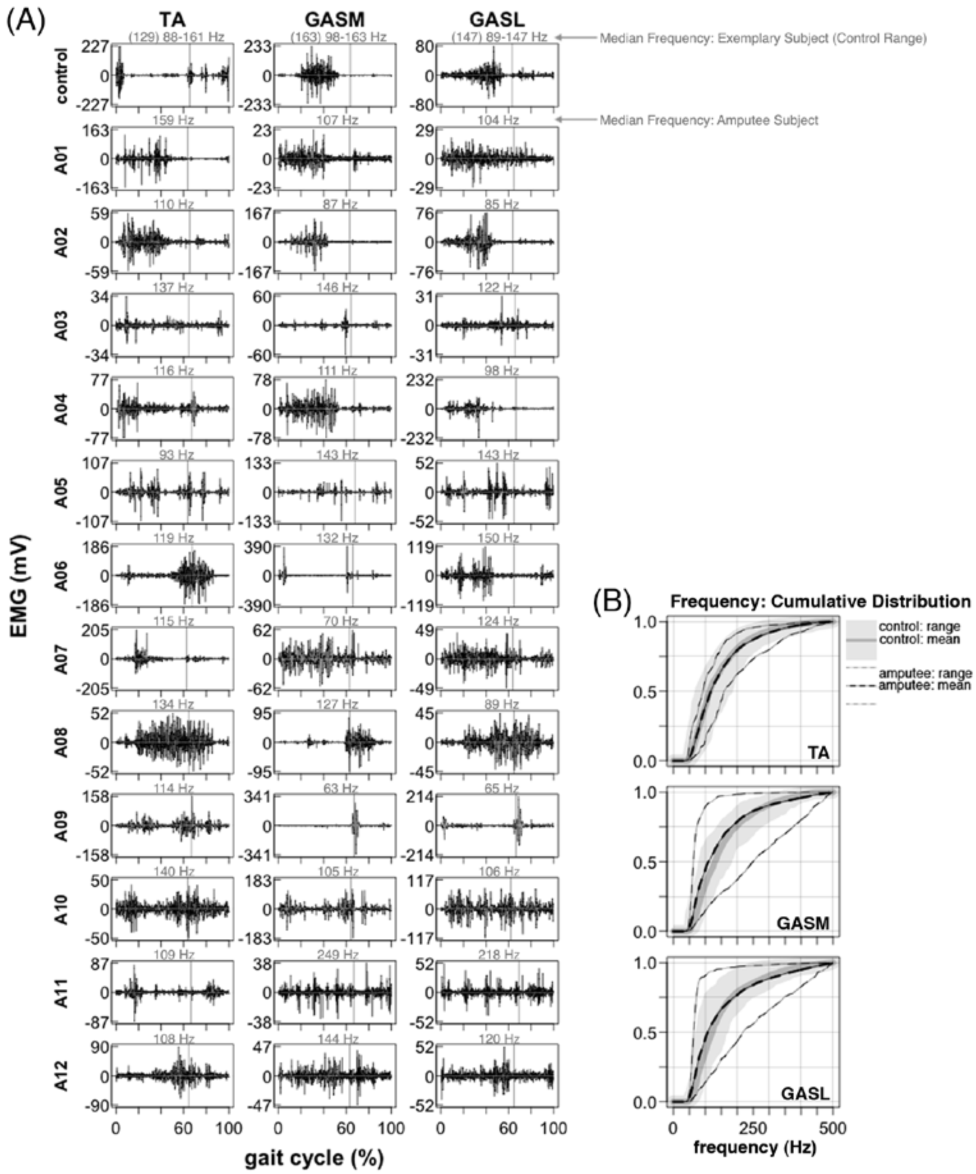

Figure 2-3. Lower Leg EMG Activation During 1.0 m/s Walking.
Tibialis Anterior (TA), *Gastrocnemius Medial Head* (GASM), *Gastrocnemius Lateral Head* (GASL). (A) Raw EMG signals from the *tibialis anterior* and *gastrocnemii* muscles for a single stride (1.0 m/s). Data is shown for one exemplary control subject and twelve amputee subjects. EMG signals are high-pass filtered and demeaned. Vertical lines show toe-off. Median frequency is shown above each plot in gray. There was a lot of variability in EMG signal patterns across amputee subjects. Amputee subject *A11* (GASM, GASL) had several EMG bursts that were approximately equally spaced and of similar amplitude across the gait cycle. A similar pattern was seen in *A10* (GASL) and *A05* (TA). Amputee subject *A09* (GASM, GASL) had short EMG bursts of high amplitude that occurred shortly after toe-off. A similar pattern was seen in *A06* (GASM) with two high-amplitude EMG bursts that occurred shortly after heel-strike and shorty before toe-off. In both *A06* and *A09,* the amplitude of the EMG bursts exceeded those recorded during maximum activation trials. (B) Empirical cumulative density function of EMG power spectrum. Lines shown for group means, boundaries indicate group range.

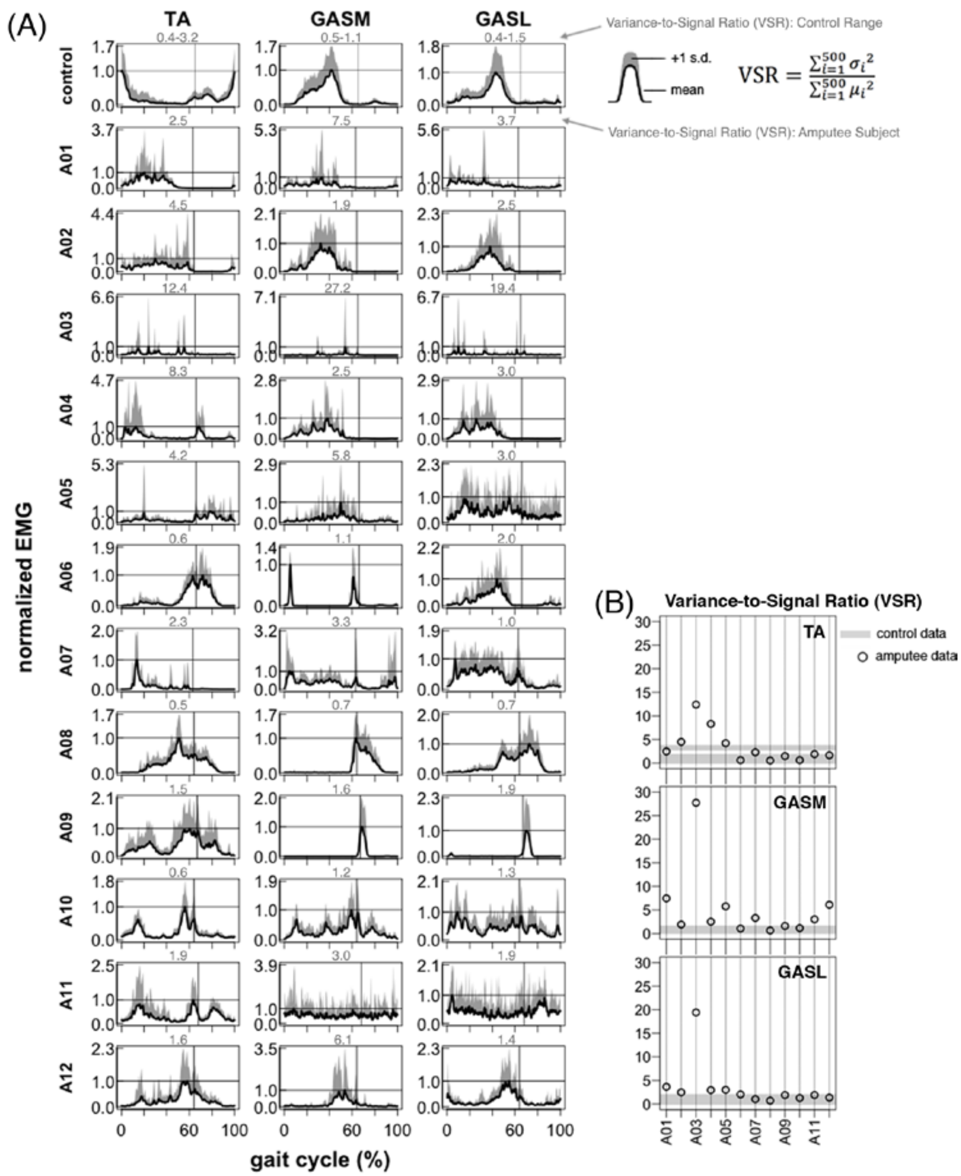

Figure 2-4. Lower Leg EMG Activation Profiles During 1.0 m/s Walking.
Tibialis Anterior (TA), *Gastrocnemius Medial Head* (GASM), *Gastrocnemius Lateral Head* (GASL). (A) Normalized mean EMG intensity curves for the *tibialis anterior* and *gastrocnemii* muscles calculated from forty consecutive strides (1.0 m/s). Control data is the grand mean of twelve control subjects. Maximum mean EMG intensity across the gait cycle is 1.0. One standard deviation above the mean is shown in gray. Vertical lines show average toe-off. Variance-to-signal ratio is shown above each plot in gray. (B) Variance-to-signal ratio of lower leg muscles calculated from 40 consecutive cycles at 1.0 m/s.

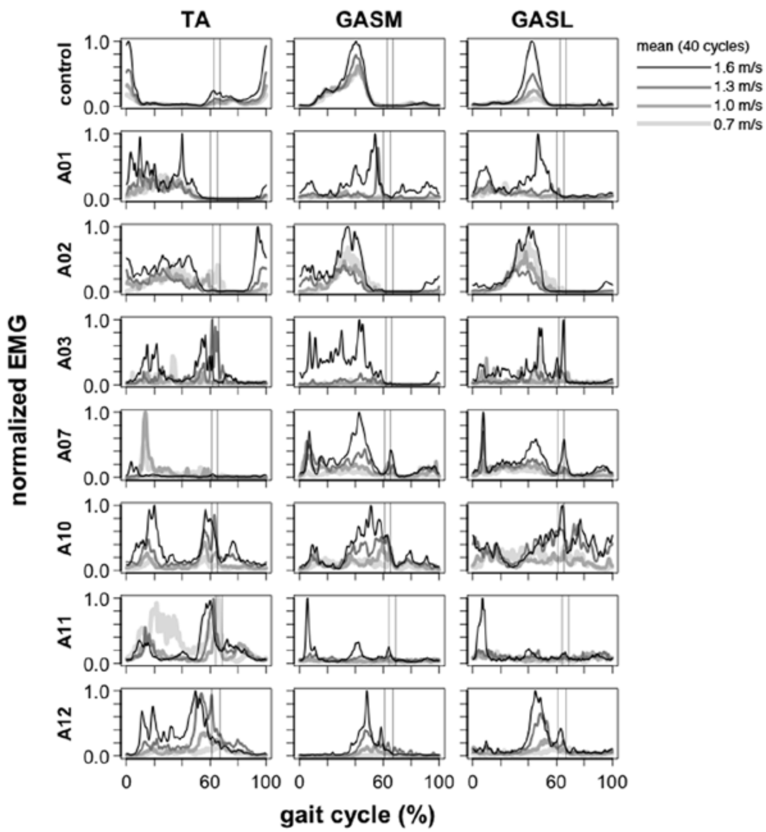

Figure 2-5. Lower Leg EMG Activation Profiles During 0.7-1.6 m/s Walking.
Tibialis Anterior (TA), *Gastrocnemius Medial Head* (GASM), *Gastrocnemius Lateral Head* (GASL). Mean EMG intensity curves of lower leg muscles for control group and seven amputee subjects during 0.7, 1.0, 1.3, and 1.6 m/s treadmill walking. Mean curves are calculated from 40 consecutive cycles. The grand mean curve is shown for the control group. Vertical lines show average toe-off events for the fastest and slowest walking speeds. In amputee subjects, the trend of increasing EMG amplitude with walking speed was not seen across amputee subjects. In amputee subject *A02*, the TA amplitude at 80-100% gait cycle scaled with speed and the GASM/GASL amplitude decreased with speed from 0.7-1.3 m/s then increased at 1.6 m/s. In subject *A07*, the TA at 0-20% gait cycle had relatively low activation higher speeds and high activation at 0.7-1.0 m/s. A similar pattern was seen in A12 with very high activation of the TA at 20-40% gait cycle at the slowest speed and relatively low activation at 0.7-1.3 m/s. In subject *A11*, the GASM/GASL at 0-20% of the gait cycle had relatively low activation at 0.7-1.3 m/s, but had large increase in amplitude at 1.6 m/s. In subject *A12*, there was a phase shift and increase in amplitude with speed for the TA and GASM/GASL at 40-60% gait cycle.

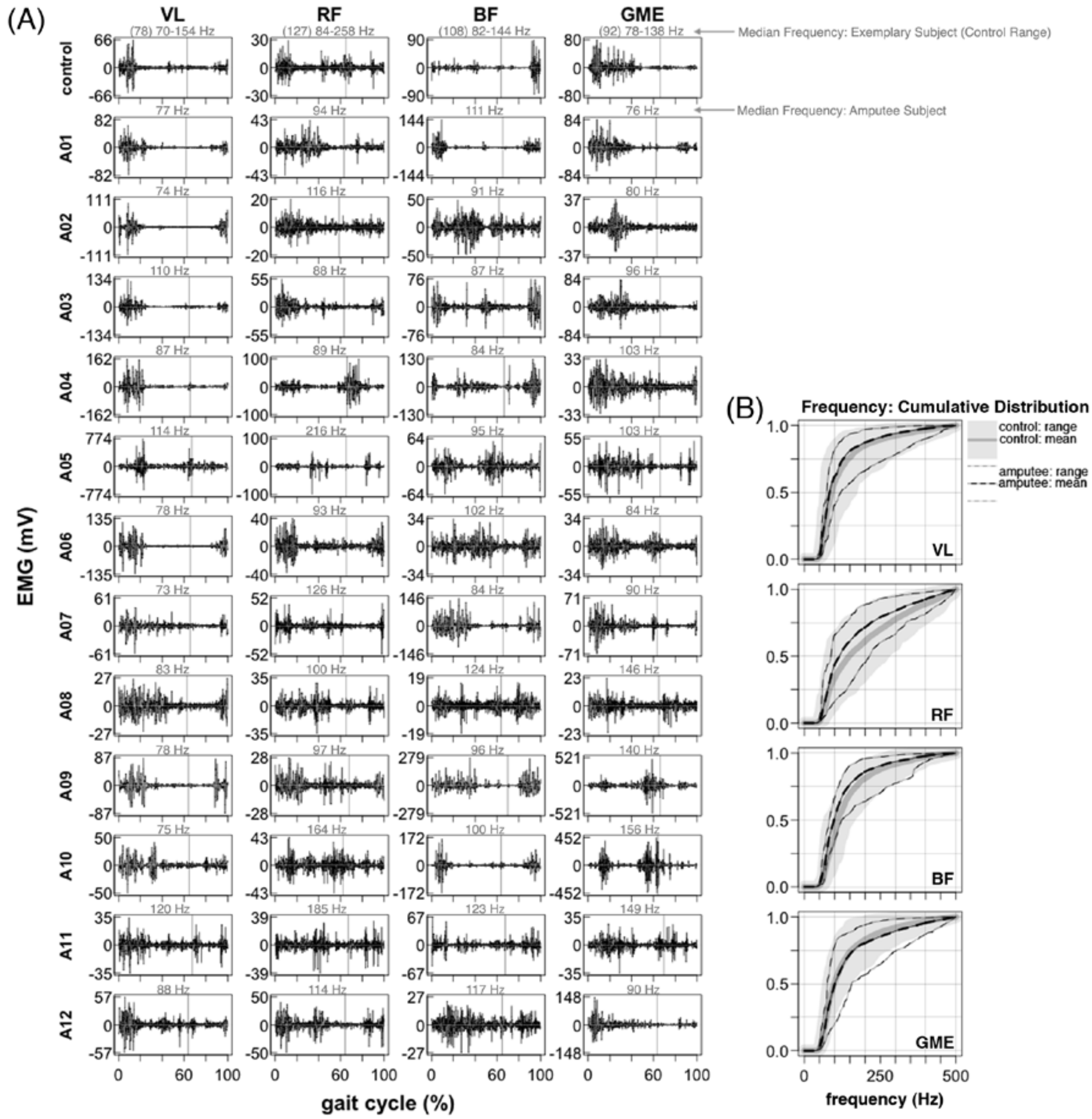

Figure 2-6. Upper Leg EMG Activation During 1.0 m/s Walking.
Vastus Lateralis (VL), *Rectus Femoris* (RF), *Biceps Femoris* (BF), *Gluteus Medius* (GME). Raw EMG signals from the *vastus lateralis*, *rectus femoris*, *biceps femoris*, *and gluteus medius* muscles for a single stride (1.0 m/s). Data is shown for one exemplary control subject and twelve amputee subjects. EMG signals are high-pass filtered and demeaned. Vertical lines show toe-off. Median frequency is shown above each plot in gray. Many EMG patterns of amputee subjects are different from the control and there is a large amount of variability in EMG patterns across amputees. (B) Empirical cumulative density function of EMG power spectrum. Lines are shown for group means and boundaries indicate group range.

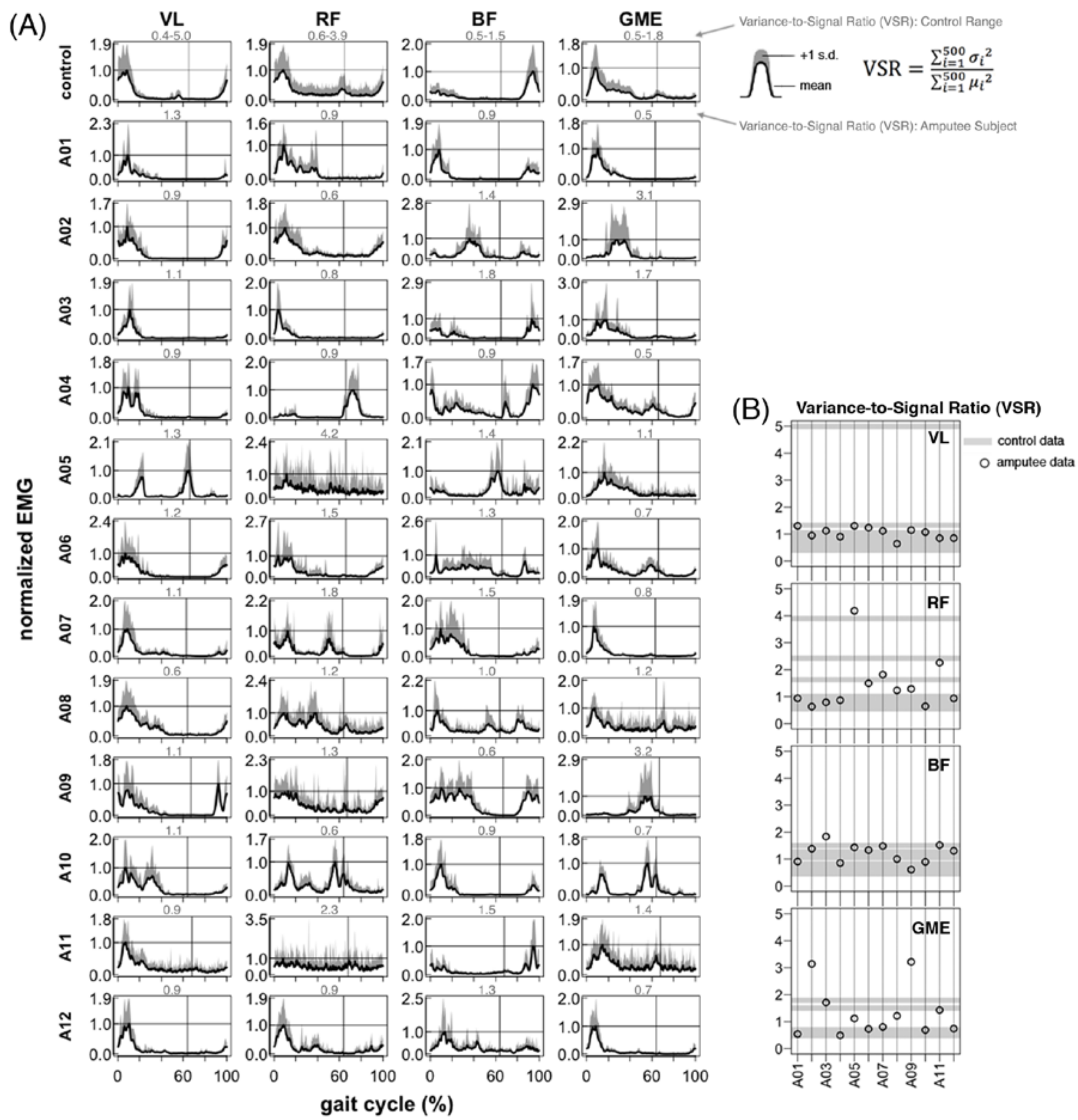

Figure 2-7. Upper Leg EMG Activation Profiles During 1.0 m/s Walking.
Vastus Lateralis (VL), *Rectus Femoris* (RF), *Biceps Femoris* (BF), *Gluteus Medius* (GME). (A) Normalized mean EMG intensity curves for the *vastus lateralis*, *rectus femoris*, *biceps femoris*, and *gluteous medius* muscles calculated from forty consecutive strides (1.0 m/s). Control data is the grand mean of twelve control subjects. Maximum mean EMG intensity across the gait cycle is 1.0. One standard deviation above the mean is shown in gray. Vertical lines show average toe-off. Variance-to-signal ratio is shown above each plot in gray. (B) Variance-to-signal ratio of lower leg muscles calculated from 40 consecutive cycles at 1.0 m/s.

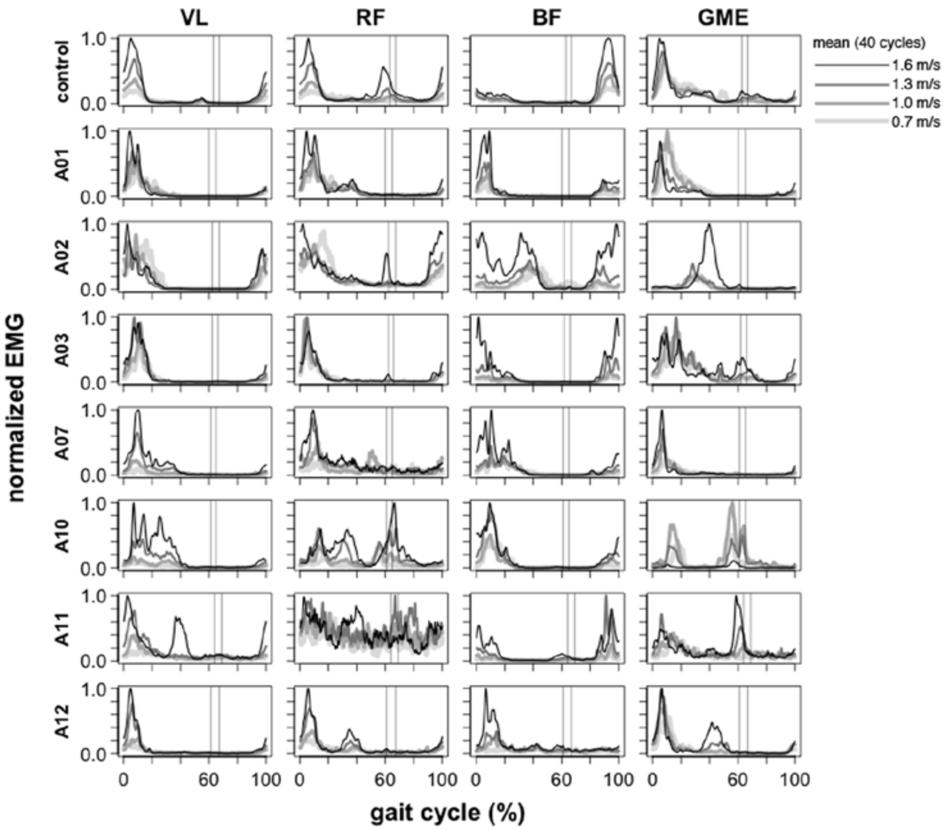

Figure 2-8. Upper Leg EMG Activation Profiles During 0.7-1.6 m/s Walking.
Vastus Lateralis (VL), *Rectus Femoris* (RF), *Biceps Femoris* (BF), *Gluteus Medius* (GME). Mean EMG intensity curves of upper leg muscles for control group and seven amputee subjects during 0.7, 1.0, 1.3, and 1.6 m/s treadmill walking. Mean curves are calculated from 40 consecutive cycles. The grand mean curve is shown for the control group. Vertical lines show average toe-off events for the fastest and slowest walking speeds. In amputee subjects, the trend of increasing EMG amplitude with walking speed was not seen across amputee subjects. In amputee subject *A11*, activation of the VL increased with walking speed at 0-20% of the gait cycle and also a phase shift (max activation appears to occur earlier). There was also activation of the VL around 40% of the gait cycle, but only at the fastest walking speed. There was no distinct activation pattern of the RF at any speed. There was GME activation around 60% of the gait cycle and amplitude increased with walking speed and also a phase shift (max activation appears to occur earlier). In subject *A10*, GME activation decreased with walking speed at 0-20% and 40-80% of the gait cycle. In subject *A03*, there was similar activation of the VL and RD across all walking speeds. In subject *A02*, activation of GME increased dramatically at 20-60% gait cycle for the fastest walking speed with a significant phase shift (peak activation occurs later). There was also a large increase in BF activation at the fastest walking speed.

Mean cross-correlation for individual amputee EMG patterns vs. the amputee mean $(\rho_{\bar{Y}Y_i})$ ranged from 0.35-0.72 for the *biceps femoris* and *gluteus medius*. In comparison, mean cross-correlation for individual control EMG patterns vs. the control mean $(\rho_{\bar{X}X_i})$ ranged from 0.72- 0.89 for the same muscles (Table 2-2). There was no significant difference (post-hoc *t*-test p>0.05) in EMG activation shape between amputees and controls for the *vastus lateralis*, *rectus femoris*, and *gluteus medius* $(\rho_{\bar{X}Y_i}, \rho_{\bar{X}X_i})$ (Table 2-2). Mean cross-correlation for individual amputee EMG patterns vs. the amputee mean $(\rho_{\bar{X}Y_i})$ ranged from 0.50-0.84 for the *vastus lateralis*, *rectus femoris*, and *gluteus medius* (Table 2-2). Mean cross-correlation for individual control EMG patterns vs. the control mean $(\rho_{\bar{X}X_i})$ ranged from 0.63-0.90 for the same muscles (Table 2-2). However, the EMG activation shape for the *biceps femoris* was significantly different between the amputee subjects and the control subjects (post-hoc *t*-test p<0.001). Mean cross-correlation for individual amputee EMG patterns against the control mean $(\rho_{\bar{X}Y_i})$ ranged from 0.31-0.38 for the *biceps femoris* (Table 2-2). Mean cross-correlation for individual control EMG patterns against the control mean $(\rho_{\bar{X}X_i})$ ranged from 0.75-0.89 for the same muscle (Table 2-2).

Table 2-2. EMG Activation Pattern Cross-Correlations

0.7 m/s	$\rho_{\bar{X}X_i}$ mean (sd)	$\rho_{\bar{X}Y_i}$ mean (sd)	$\rho_{\bar{Y}Y_i}$ mean (sd)
Tibialis Anterior	0.73 (0.11) *°	–0.33 (0.13) °	0.44 (0.21) *
Gastrocnemius Medial Head	0.90 (0.09) *°	0.48 (0.42) °	0.45 (0.32) *
Gastrocnemius Lateral Head	0.79 (0.17) *°	0.37 (0.40) °	0.37 (0.35) *
Vastus Lateralis	0.81 (0.19)	0.83 (0.08)	0.89 (0.07)
Rectus Femoris	0.63 (0.28)	0.70 (0.23)	0.71 (0.18)
Biceps Femoris	0.75 (0.10) *°	0.31 (0.48) °	0.35 (0.36) *
Gluteus Medius	0.72 (0.31) *	0.66 (0.32)	0.67 (0.28) *
1.0 m/s	$\rho_{\bar{X}X_i}$ mean (sd)	$\rho_{\bar{X}Y_i}$ mean (sd)	$\rho_{\bar{Y}Y_i}$ mean (sd)
Tibialis Anterior	0.80 (0.09) *°	–0.24 (0.08) °	0.32 (0.25) *
Gastrocnemius Medial Head	0.87 (0.08) *°	0.23 (0.40) °	0.20 (0.19) *
Gastrocnemius Lateral Head	0.83 (0.12) *°	0.20 (0.37) °	0.24 (0.37) *
Vastus Lateralis	0.86 (0.08)	0.83 (0.10)	0.90 (0.06)
Rectus Femoris	0.77 (0.16)	0.70 (0.23)	0.70 (0.23)
Biceps Femoris	0.86 (0.06) *°	0.38 (0.39) °	0.53 (0.30) *
Gluteus Medius	0.82 (0.14) *	0.63 (0.36)	0.61 (0.32) *
1.3 m/s	$\rho_{\bar{X}X_i}$ mean (sd)	$\rho_{\bar{X}Y_i}$ mean (sd)	$\rho_{\bar{Y}Y_i}$ mean (sd)
Tibialis Anterior	0.84 (0.08) *°	–0.09 (0.18) °	0.20 (0.22) *
Gastrocnemius Medial Head	0.88 (0.07) *°	0.32 (0.26) °	0.41 (0.24) *
Gastrocnemius Lateral Head	0.92 (0.07) *°	0.32 (0.26) °	0.22 (0.32) *
Vastus Lateralis	0.89 (0.05)	0.84 (0.11)	0.84 (0.11)
Rectus Femoris	0.82 (0.15)	0.70 (0.23)	0.70 (0.23)
Biceps Femoris	0.89 (0.05) *°	0.33 (0.34) °	0.55 (0.26) *
Gluteus Medius	0.77 (0.19) *	0.63 (0.38)	0.56 (0.35) *
1.6 m/s	$\rho_{\bar{X}X_i}$ mean (sd)	$\rho_{\bar{X}Y_i}$ mean (sd)	$\rho_{\bar{Y}Y_i}$ mean (sd)
Tibialis Anterior	0.85 (0.07) *°	–0.05 (0.36) °	0.27 (0.28) *
Gastrocnemius Medial Head	0.88 (0.07) *°	0.48 (0.28) °	0.53 (0.15) *
Gastrocnemius Lateral Head	0.91 (0.09) *°	0.40 (0.40) °	0.46 (0.34) *
Vastus Lateralis	0.90 (0.06)	0.77 (0.20)	0.74 (0.15)
Rectus Femoris	0.74 (0.15)	0.58 (0.30)	0.66 (0.33)
Biceps Femoris	0.89 (0.07) *°	0.31 (0.26) °	0.72 (0.13) *
Gluteus Medius	0.75 (0.18) *	0.50 (0.40)	0.45 (0.30) *

X=controls, Y=amputees; *p<0.001, $\rho_{\bar{X}X_i}$ vs. $\rho_{\bar{Y}Y_i}$; °p<0.001, $\rho_{\bar{X}X_i}$ vs. $\rho_{\bar{X}Y_i}$

Inter-Stride Variability of EMG During Walking

Variance-to-signal ratios of EMG during 1.0 m/s treadmill walking were significantly greater in the amputee group compared to the control group (control mean=1.0, amputee mean=2.4; ANOVA group effect, $p<0.001$) (Table 2-3, Figures 2-4 and 2-7). However, post-hoc t-tests revealed that the only muscle with a significant difference between groups was the *gastrocnemius medial head* (post-hoc *t*-test $p<0.001$).

Table 2-3. Variance-to-Signal Ratios, 1.0 m/s Treadmill Walking

Muscle	Controls mean (sd)	Amputees mean (sd)
Tibialis Anterior	1.0 (0.8)	3.4 (3.6)
Gastrocnemius Medial Head	0.8 (0.2) *	5.2 (7.4) *
Gastrocnemius Lateral Head	0.9 (0.3)	3.5 (5.1)
Vastus Lateralis	1.1 (1.2)	1.0 (0.2)
Rectus Femoris	1.2 (1.0)	1.4 (1.0)
Biceps Femoris	0.9 (0.3)	1.2 (0.4)
Gluteus Medius	0.7 (0.4)	1.3 (0.9)

*$p<0.001$, controls vs. amputees

EMG Median Frequencies

During maximum voluntary activation, median EMG frequencies for lower leg muscles were significantly lower in amputee subjects compared to control subjects (ANOVA group effect, $p<0.001$) (Table 2-4, Figure 2-2B). However, during 1.0 m/s treadmill walking, median EMG frequencies for upper and lower leg muscles of amputee and control subjects were not significantly different (ANOVA group effect, $p>0.10$) (Table 2-4, Figures 2-3 and 2-6). In the amputee group, median EMG frequencies of residual lower leg muscles were similar for maximum voluntary activation and 1.0 m/s treadmill walking (post-hoc *t*-test, $p>0.50$) (Table 2-4). In the control group, median EMG frequencies of lower leg muscles were significantly greater during maximum voluntary activation compared to 1.0 m/s treadmill walking (post-hoc *t*-test, $p<0.001$) (Table 2-4).

Table 2-4. EMG Median Frequencies

	Maximum Voluntary Activation		Treadmill Walking (1.0 m/s)	
Muscle	Controls mean (sd)	Amputees mean (sd)	Controls mean (sd)	Amputees mean (sd)
Tibialis Anterior	153 (14) *	127 (23) *	115 (22)	121 (18)
Gastrocnemius Medial Head	174 (23) *	137 (26) *	131 (18)	124 (49)
Gastrocnemius Lateral Head	166 (23) *	124 (34) *	122 (14)	119 (40)
Vastus Lateralis			97 (24)	88 (17)
Rectus Femoris			156 (62)	123 (42)
Biceps Femoris			113 (18)	101 (15)
Gluteus Medius			102 (18)	109 (30)

*p<0.001, controls vs. amputees; p<0.001, maximum voluntary activation vs. treadmill walking

DISCUSSION

The main finding of this study is that during walking, most amputee subjects had residual lower leg muscle activation patterns that were entrained to the gait cycle but highly variable across subjects. The residual lower leg muscle activation patterns were very different from the normal control patterns (Figure 2-4). This is evidenced by the low EMG cross-correlation values between amputee subjects and the control mean for *tibialis anterior* and *gastrocnemii* (Table 2-2). Despite the high variability in residual lower leg EMG patterns across amputee subjects, inter-stride variability was similar to that of control subjects. The *gastrocnemius medial head* was the only muscle with a variance-to-signal ratio significantly greater in the amputee group compared to the control group. This significant difference in variance-to-signal ratio between groups was due to a single amputee subject whose variance-to-signal noise ratio was magnitudes greater than other amputee subjects (Figure 2-4, subject *A03*). Subject *A03* had high inter-stride variability for all three residual lower leg muscles (Figure 2-4). The inter-stride variability could be problematic if it continued when using a powered lower limb prosthesis under myoelectric control. However, it seems reasonable to presume that the inter-stride variability would decrease if the residual muscle activity had a functional purpose during walking (e.g., to control dynamics of a powered prosthesis). Future studies should document the variability in muscle recruitment patterns while subjects learn to use powered prostheses.

Another finding of this study is that many, but not all, amputee subjects had robust volitional control of residual lower leg muscle activation. During maximum voluntary dorsiflexion and plantar flexion, residual muscle activation profiles in several amputee subjects were similar to controls (Figure 2-2). The maximum activation levels were well above resting baseline, the time to reach maximum activation from resting baseline was short, and the activation levels were well sustained. Some of the amputee subjects were able to differentiate *tibialis anterior* and *gastrocnemii* activation and had coactivation levels similar to control subjects (e.g., Figure 2-2A, subjects *A05* and *A09*). Other amputee subjects were not able to differentiate *tibialis anterior* and *gastrocnemii* activation during volitional maximum activation. As a result, there was either complete coactivation for both plantar flexion and dorsiflexion tasks (e.g., Figure 2-2A, subject *A01*) or an inability to recruit any muscles strongly during dorsiflexion (e.g., Figure 2-2A, subjects *A02* and *A08*). For the subjects that demonstrated complete coactivation, synchronous recruitment of residual muscles was not hard-wired because their *tibialis anterior* and *gastrocnemii* activation patterns were distinctly different from each other during walking, especially at faster walking speeds (e.g., Figure 2-2A, subjects *A01* and *A02*). One reason that the amputee subjects may have lost robust volitional control of the residual limb muscles is the lack of proprioceptive or visual feedback of muscle activity. Without an ankle joint to provide sensory information about joint position, there is no clear information reinforcing the consequences of muscle activity. It seems likely that coupling a powered prosthetic limb to the residual limb muscle activity would increase the volitional motor control [27, 35-37].

In the upper leg muscles, our data show that amputee subjects had greater inter-subject variability in their *biceps femoris* and *gluteus medius* muscle activation profiles compared to control subjects during walking (Table 2-2, Figure 2-8). In addition, our data show that amputee subjects had a different *biceps femoris* activation profile shape than control subjects (Table 2-2, Figure 2-8). Previous studies have suggested that transtibial amputees walk with greater residual leg *biceps femoris* activation during early stance compared to the intact *biceps femoris* to stabilize the knee joint [38-40] and/or increase propulsion of the residual leg [41, 42]. In normal walking, the primary function of the *gluteus medius* is to provide support during early stance to midstance and the *biceps femoris* has the potential for generating support from early stance to midstance. Ankle dorsiflexors provide support during early stance and ankle plantar flexors provide support during late stance [43]. It is likely that transtibial amputees compensate for the loss of support from ankle muscles by recruiting muscles above the knee to increase walking stability during stance. The inter-subject variability in the *biceps femoris* and *gluteus medius* activation shape observed in our amputee subjects suggests that there are differences in compensatory muscle recruitment patterns used by transtibial amputees during walking.

One limitation of our study is that we did not present data from overground walking. Past studies have shown that lower limb EMG patterns and kinematics can be different during treadmill walking compared to overground walking [44, 45]. Biomechanically, treadmill gait and overground gait is identical if the treadmill belt speed is constant [46]. The differences in biological gait measurements occur primarily due to two aspects: differences in visual flow [47] and treadmill speed fluctuations [48]. We did not include overground walking in this study because our primary focus was to quantify differences in signal patterns and variability between amputee and non-amputee groups and within groups. Now that we have demonstrated that reliable signals can be recorded from residual muscles of transtibial amputees during treadmill walking at constant speeds, we plan to expand our study to include lower limb EMG patterns of transtibial amputees and non-amputees during overground walking at self-selected walking speeds. This will provide a better understanding of how signals recorded from residual muscles in transtibial amputees can be utilized to control robotic lower limb prostheses. Another limitation of our study is that the mean age of our amputee group was greater than our non-amputee group. We do not believe that the results presented in this study would change significantly given more similar ages between groups, but further data could support or refute this assumption.

Several previous studies have presented EMG data from the amputated limb of transtibial amputees during walking [38, 39, 41, 49], but they did not record EMG from residual limb muscles inside the socket. It has traditionally been thought that the mechanics of the socket-limb interface prevent reliable measurements of EMG from the residual limb muscles during walking with surface electrodes. Au et al. recorded EMG from residual limb muscles within the socket, but were only able to get a reliable signal during swing [6]. We were able to record robust and reliable EMG during both stance and swing by using active EMG electrodes to maximize signal-to-noise ratio and using silicone putty to minimize movement and discomfort at the electrode sites.

Although there was the possibility for mechanical artifacts in our EMG recordings, data of EMG median frequencies suggest that we measured muscle activity from the residual limb muscles with little to no motion artifact. The EMG median frequencies recorded from the residual limb muscles during walking were similar to the EMG median frequencies recorded from the residual limb muscles during seated maximum voluntary activation trials (Table 2-4). In addition, the EMG median frequencies recorded from residual lower leg muscles in amputee subjects during treadmill walking were similar to the EMG median frequencies of the intact lower leg muscles in control subjects during treadmill walking (Table 2-4). Some of the amputee subjects demonstrated abnormal EMG patterns that had rhythmic, short-duration, and high-amplitude bursts (e.g., Figure 2-3, subjects *A06* and *A09*). We do not believe that these bursts resulted from mechanical perturbations to the electrodes because of the filtering we used and the frequency content of the resulting signals. Similar EMG patterns have been demonstrated in individuals with spinal cord injury that have had long-term disuse atrophy of the muscles [50, 51]. The short-duration, high-amplitude EMG bursts that occurred around heel-strike and toe-off events may have been a result of reflex activation from muscle fiber stretch (Ia and II afferents) or rapid loading/unloading (Ib afferents).

The unique residual muscle activation patterns seen in our amputee subjects during gait suggest that neural plasticity may have occurred following amputation. Previous studies have demonstrated that neural plasticity in lower limb amputees occurs predominantly at the cortical level [52, 53]. Neural plasticity can be affected by cause of amputation (e.g. traumatic, cancer-related, dysvascular-related), age at amputation, surgical procedure, muscle atrophy, and degeneration of nerves. The long-term cortical reorganization that occurs following injury is also highly use-dependent [54]. Changes in gait-related muscle activity following amputation would have a major impact on use-dependent cortical plasticity. Some amputees may learn to activate their residual muscles to improve stability at the limb-socket interface or to minimize socket discomfort/pain associated with impulsive prosthetic forces. This could alter the activation patterns away from the normal functional pattern seen in intact subjects and could contribute to increased inter-subject variability in amputees.

The results of this study are encouraging for the development of powered lower limb prosthesis under myoelectric control. Coupling an amputee's nervous system to a robotic prosthesis should provide a strong stimulus for learning to modify residual muscle activation patterns. In past studies, we have found that subjects with intact musculoskeletal systems can quickly adapt their muscle activation patterns to control powered lower-limb orthoses under proportional myoelectric control [55-58]. It seems likely that amputees could also learn to modify their muscle activation patterns to control powered lower-limb prostheses, though it may take longer due to the motor plasticity that has occurred since the amputation. Residual limb muscle activation patterns during dynamic tasks such as walking may function to improve fit and/or minimize discomfort at the socket-limb interface. Learning new residual activation patterns to control lower-limb prostheses may compete with this. Future studies should investigate why amputees adopt specific residual limb muscle activation patterns in order to assess the feasibility of myoelectric control using residual limb muscles during walking. Continued technological advances in intramuscular electrodes that could transmit control EMG signals through the prosthetic socket-limb interface without breaking the skin [28-30] would provide a means for generating feedforward control signals to a robotic prosthesis from the nervous system. Another option is recent technological advances in flexible epidermal electronics that could be mounted directly on the skin within the prosthetic socket-limb interface [59]. Either of these options could provide a long-term means for improving the control of powered lower limb prosthesis using EMG from the residual limb muscles.

CONCLUSIONS

It is possible to record artifact-free muscle activation patterns from residual limb muscles within the prosthetic socket-limb interface with surface electromyography electrodes. There is high inter-subject variability in recruitment patterns in amputees, but for each subject EMG patterns are consistent from stride to stride. Our results support the potential use of myoelectric controllers for direct feedforward control of robotic lower limb prostheses.

Chapter 3: An Experimental Powered Lower Limb Prosthesis Using Proportional Myoelectric Control

This chapter has been previously published:
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ABSTRACT

One way to provide powered lower limb prostheses with greater adaptability to a wearer's intent is to use a neural signal to provide feedforward control of prosthesis mechanics. We designed and tested the feasibility of an experimental powered ankle-foot prosthesis that uses pneumatic artificial muscles and proportional myoelectric control to vary ankle mechanics during walking. The force output of the artificial plantar flexor muscles was directly proportional to the subject's residual gastrocnemius muscle activity. The maximum force generated by a pair of artificial muscles fixed at nominal length was 3513 N. The maximum planter flexion torque that could be generated during walking was 176 Nm. The force bandwidth of the pneumatic artificial muscles was 2 Hz. The electromechanical delay was 33 ms, the time to peak tension was 48 ms, and the half relaxation time was 50 ms. We used two artificial muscles as

dorsiflexors and two artificial muscles as plantar flexors. The prosthetic ankle had 25 degrees of dorsiflexion and 35 degrees of plantar flexion with the artificial muscles uninflated. The intent of the device was not to create a commercially viable prosthesis, but to have a laboratory prototype to test principles of locomotor adaptation and biomechanics. We recruited one unilateral transtibial amputee to walk on a treadmill at 1.0 m/s while wearing the powered prosthesis. We recorded muscle activity within the subject's prescribed prosthetic socket using surface electrodes. The controller was active throughout the entire gait cycle and did not rely on detection of gait phases. The amputee subject quickly adapted to the powered prosthesis and walked with a functional gait. The subject generated peak ankle power at push off that was similar between amputated and prosthetic sides. Our results suggest that amputees can use their residual muscles for proportional myoelectric control to alter prosthetic mechanics during walking.

INTRODUCTION

Although robotic technologies have made powered lower limb prostheses technically viable for the first time, there is still a need for more adaptable and functional controllers. Clinically available powered lower limb prostheses generally rely on intrinsic sensing of kinematics and kinetics to control the mechanics of the actuators via state-based control [6, 12, 60]. Clinicians can tune the prosthesis controller to each individual patient to maximize prosthesis performance, but there are limits as to how well the prosthesis adapts to different terrain (e.g. sandy beach) and locomotor tasks (e.g. weaving through a crowded subway station). One way to give amputees greater volitional and active control of a powered prosthesis is proportional myoelectric control [61-64]. If lower limb amputees can actively change the prosthetic ankle mechanics in real time in response to a feedforward neural signal, it would give them the opportunity to learn and adapt to a range of locomotor challenges.

Transtibial amputees generally have robust and reliable residual muscle activation patterns that might provide a source for proportional myoelectric control [61-64]. In our past study, we placed electromyography sensors inside the socket-limb interface for twelve amputee subjects as they walked on a treadmill. Although there was considerable variability in muscle activity patterns across amputee subjects, variance-to-signal noise ratios and step-to-step variability of the amputee subjects were similar to those measured in intact subjects. This suggests that amputees had learned a new muscle activation pattern compared to their intact gait, but that the pattern was fairly reliable.

The purpose of this study is to design an experimental powered prosthesis that uses pneumatic artificial muscles and proportional myoelectric control. Our goal was not to build a commercially available prosthesis but to design a research prototype that could be used to examine the viability of proportional myoelectric control and other control schemes. The pneumatic artificial muscles have inherent compliance, high mechanical power capabilities, along with other mechanical properties advantageous to use with biorobotics [65-67]. They do depend on a compressed air source, but given our intended use in a laboratory for controller testing, the need for compressed air was not a major disadvantage for the prototype.

METHODS

Device

We fabricated an experimental powered prosthesis with an interchangeable prosthetic socket (Figure 3-1) at the University of Michigan Orthotics and Prosthetics Center and the University of Michigan Human Neuromechanics Laboratory. We used a freely articulating ankle (Rampro, Oceanside, CA) and a low-profile flex-foot (Trulife, Dublin, Ireland). We used two pneumatic artificial muscles as dorsiflexors and two pneumatic artificial muscles as plantar flexors. The dorsiflexor and plantar flexor moment arms were 5 cm. We used a pair of pneumatic artificial muscles for dorsiflexors and plantar flexors so that we could easily adjust ankle stiffness and maximum plantar flexion torque. The mass of powered prosthesis without a prosthetic socket was 2.8 kg. Although the mass of the powered prosthesis was heavier than the subject's prescribed prosthesis, the subject's pin-lock suspension was sufficient for walking with the powered prosthesis. While walking with the powered prosthesis, the subject indicated that he did not notice an increase in pistoning over his prescribed prosthesis.

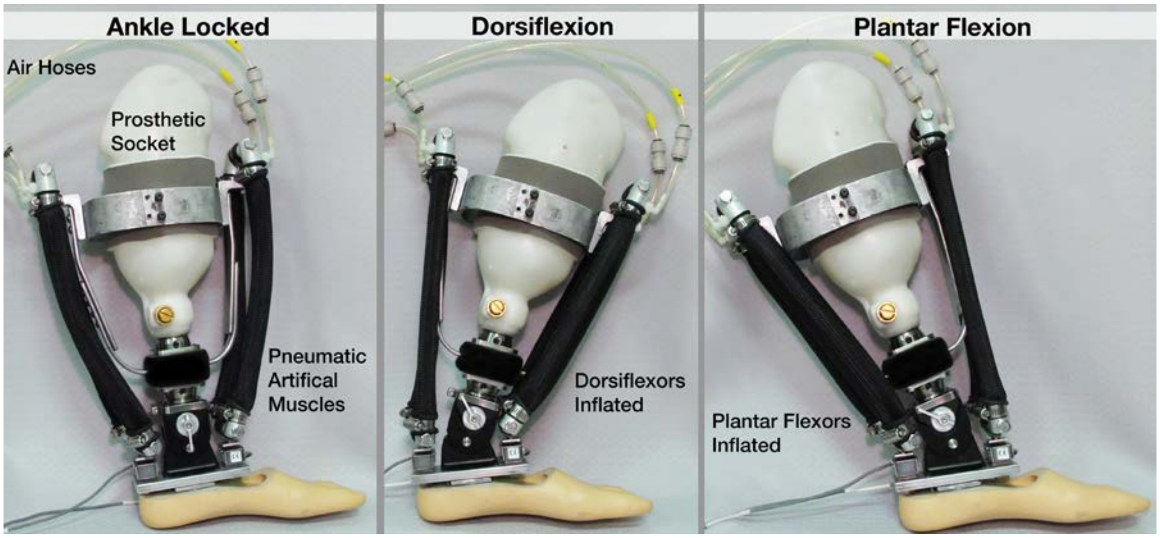

Figure 3-1. Experimental Powered Prosthesis (Pneumatic Artificial Muscles).
The powered prosthesis in three different ankle positions. Two artificial muscles acted as doriflexors and two artificial muscles acted as plantar flexors. The range of motion of the ankle with uninflated actuators was 25 degrees of dorsiflexion and 35 degrees of plantar flexion. The prosthetic socket was interchangeable so that the amputee's prescribed socket could be used. Standard stainless steel prosthetic components above the ankle and below the socket interface allowed for proper alignment. The ankle was a modified Rampro Swim Ankle (Rampro, Oceanside, CA). The foot was a modified Seattle LiteFoot (Trulife, Dublin, Ireland).

Controller

To control inflation of the pneumatic artificial muscles during bench-top testing and subject testing, we connected two pressure regulators (MAC Valves, Wixom, MI) to each artificial muscle (maximum capacity of 90 psi - 6.2 bar). The input signal for each pressure regulator ranged from 0 volts to 10 volts. We interfaced a desktop computer and a real-time controller board (dSPACE, Inc., Northville, MI) to generate and send input signals to the pressure regulators. We used MATLAB/Simulink running on the desktop computer to program the controller.

During walking, the control signal sent to the pressure regulators for the artificial dorsiflexor muscles was constant. The control signal sent to the pressure regulators for the artificial plantar flexor muscles was proportional to the smoothed residual gastrocnemius electromyography (EMG) signal. To obtain the smoothed EMG signal, we first attenuated motion artifact by high-pass filtering the raw EMG signal (second order Butterworth filter, 50 Hz cut-off frequency). A 50-Hz cut-off frequency for the high-pass filter was necessary to sufficiently attenuate motion artifact generated at the skin-socket interface. We then calculated a linear envelop by full-wave rectifying the signal followed by low-pass filtering the signal (second order Butterworth filter, 6 Hz cut-off frequency).

Bench-Top Testing

We calculated the electromechanical response times, force-tension curves, and force bandwidth and phase lag of the artificial pneumatic muscles using an isometric bench-top testing setup. We used a tension load cell (Omega, Stamford, CT) to measure the force produced by the pneumatic artificial muscles. We recorded load cell data at 1000 Hz using a Vicon Motion Capture system (Vicon, Lake Forest, CA).

To measure electromechanical response times, we fixed the artificial muscles at nominal length (27.5 cm) and input a 5-ms square pulse with to the proportional controller and recorded the muscle force output. The gain of the proportional controller was adjusted so that the control signal peak was just below 10 V. We computed the electromechanical delay (EMD), time to peak tension (TPT), and half relaxation time (HRT) of the pneumatic artificial muscles. We defined electromechanical delay as the time from the delivery of the square pulse to the initial rise of force generation. We defined force initiation as the time when force exceeded 3 standard deviations above baseline mean. We defined time to peak tension as the time from initial rise of force generation to the peak force. We defined half relaxation time as the time from peak force to the time where force dropped below 50 percent of peak force.

To measure force tension curves, we fixed the artificial muscles at five lengths (20, 22, 24, 26, 27.5 cm). At each length, we input four control signals (2.5, 5, 7.5, 10 V). We recorded the muscle force at each condition to generate four tension-length curves corresponding to 25, 50, 75, and 100% of maximum pressure.

To measure the bandwidth and phase lag of the muscles, we input a sinusoid signal directly to the pressure regulators at 17 frequencies between 0.5 and 6.0 Hz. The sinusoid signal had a 5-volt offset with a 10-volt range so that the input signal spanned the entire range of the input to the pressure regulators (0-10 V). We used a magnitude drop of -3dB to determine the force bandwidth of the actuator.

Subject Testing

We recruited one healthy male unilateral transtibial amputee subject (age=57 yrs, height=188 cm, mass=90 kg) to test the powered prosthesis properties during treadmill walking. The subject had his right leg amputated 3 years ago as a result of trauma. His prescribed prosthesis was the College Park TruStep (College Park, Warren, MI) with pin-lock suspension. The subject wore his prescribed liner and socket with the powered prosthesis. A certified prosthetists fit the powered prosthesis to the subject by adjusting leg length and socket alignment. We placed surface electrodes on the skin at the limb-socket interface to record the subject's residual gastrocnemius muscle activity, as described in our previous study [61]. The subject wore a bodyweight support harness for the entire testing session. The harness was not used to lift any of the subject's body weight, but to act as a safety in case of a trip or stumble. Before collecting data with the powered prosthesis, we recorded 2 minutes of walking data at 1.0 m/s with the subject's prescribed prosthesis.

We adjusted the resting activation of the pneumatic artificial muscles to provide a set-point ankle stiffness to achieve foot clearance during swing. While the subject was seated wearing the powered prosthesis, we inflated the artificial plantar flexors until there was visible tension in the muscles then inflated the artificial dorsiflexors until the ankle angle was approximately zero. We then asked the subject to stand with the powered prosthesis and adjusted the pressure in the dorsiflexors so that the foot was flat on the ground with the ankle and knee at neutral posture. To set the proportional controller gain, we asked the subject to stand with the powered prosthesis and maximally activate his residual gastrocnemius muscle so that he was standing on his toes. We increased the gain until the control signal began to saturate during standing maximum voluntary activation.

For safety and to increase the subject's confidence in the device, he practiced activating his residual gastrocnemius muscle while seated and standing to actuate the ankle and acclimate to the controller before walking with the powered prosthesis. We then had the subject walk on the treadmill with the powered prosthesis, starting at a slow speed and working up to 1.0 m/s. We had the subject continue to walk at 1.0 m/s until he indicated that he wanted to stop for a break. The subject alternated between treadmill walking trials at 1.0 m/s and rest periods for a total of 30 minutes of treadmill walking. To provide guidance during walking, we instructed the subject to relax his residual muscle during swing and begin contracting his residual muscle after heel strike. We also instructed the subject to gradually increase muscle activation from mid stance through late stance to achieve peak activation at push off.

We recorded kinematics, kinetics, and EMG data using a Vicon Motion Capture system (Vicon, Lake Forest, CA). We recorded lower body motion capture data at 100 Hz to calculate ankle joint position and ankle angle. We recorded ground reaction data at 1000 Hz using a split-belt force-measuring treadmill calculate vertical and fore-aft ground reaction forces and fore-aft center of pressure [31]. From the ankle kinematics, ground reaction forces, and center of pressure, we calculated ankle moments [68] and powers. We recorded surface EMG from the residual gastrocnemius muscle at 1000 Hz using preamplifier electrodes (Biometrics Ltd, Newport, United Kingdom).

RESULTS

Bench-Top Testing

The tension-length relationship for our pneumatic artificial muscles was approximately linear (Figure 3-2). Linearity decreased slightly with decreasing pressure. At nominal length (27.5 cm) and maximum pressure (6.2 bar), the maximum force generated by a pair of pneumatic artificial muscles was 3513 N. At nominal length and 50% of maximum pressure, the force generated by the artificial muscles was 1498 N (43% of maximum force). Assuming linearity, the force produced by the artificial muscles at the shortest functional length (23.5) and 100% of maximum pressure was 1610 N (46% of maximum force).

The force bandwidth of the pneumatic artificial muscles was 2.0 Hz (Figure 3-3). At 2.0 Hz, the phase lag was 39 degrees. At 4.0 Hz, the phase lag was 67 degrees and at 6.0 Hz, the phase lag was 81 degrees. At the lowest frequency (0.5 Hz), the phase lag was 8 degrees.

The electromechanical delay for the pair of pneumatic artificial muscles was 33 ms, the time-to-peak tension was 48 ms, and the half-relaxation time was 50 ms (Figure 3-4).

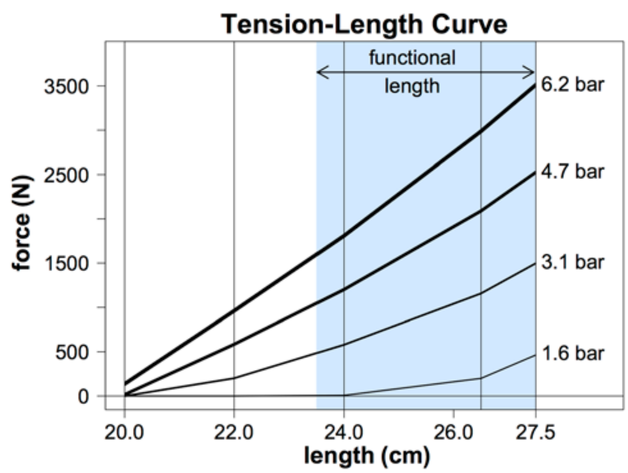

Figure 3-2. Pneumatic Artificial Muscle Tension-Length Curve.
Tension and length data from a pair of muscles during isometric bench-top testing at 20, 22, 24, 26.5, and 27.5 cm lengths and 25%, 50%, 75%, and 100% of max pressure (approximately 1.6, 3.1, 4.7, and 6.2 bar). The functional length of the artificial plantar flexor muscles during walking was 23.5-27.5 cm, which is approximately 85-100% of nominal length. We calculated he functional length as the range of muscle length measured during walking using 3-dimensional kinematics.

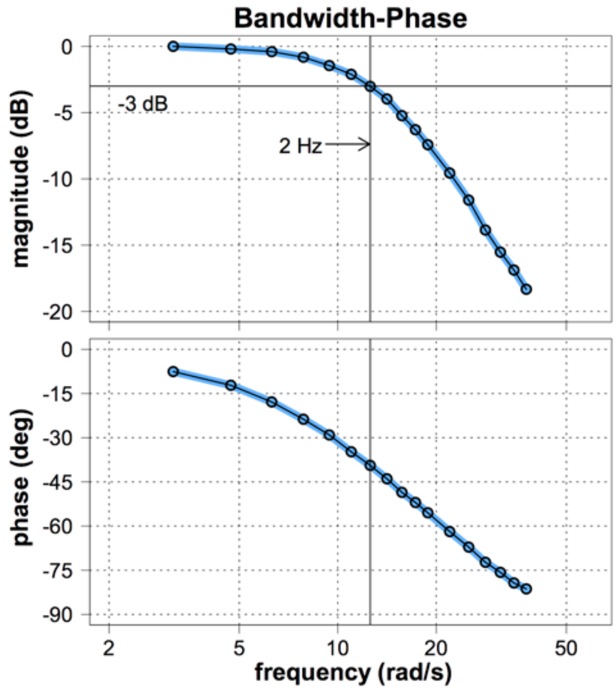

Figure 3-3. Pneumatic Artificial Muscle Force Bandwidth.
Force bandwidth and phase lag of a pair of pneumatic artificial muscles in an isometric bench-top configuration at nominal muscle length (27.5 cm). The input was a sinusoid signal with peak-to-peak amplitude of 10 volts. Point characters on the plots show the frequencies that data were recorded. The bandwidth of the artificial muscles was 2.0 Hz. At 2.0 Hz, the output lags the input by 39 degrees.

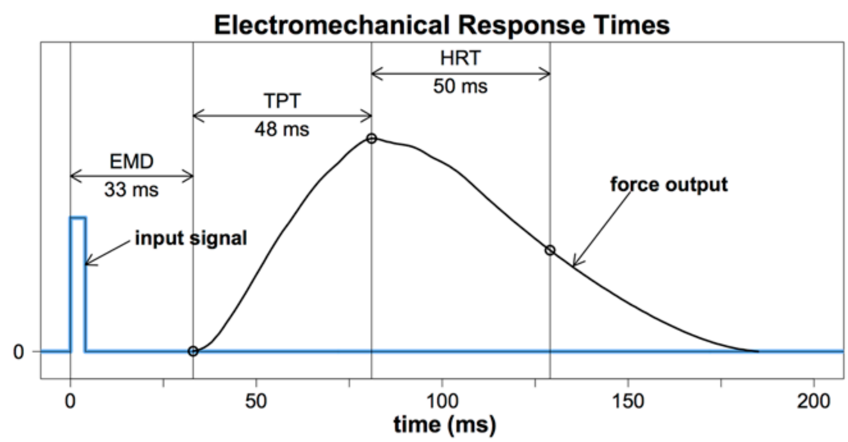

Figure 3-4. Pneumatic Artificial Muscle Response Time.
Electromechanical response times of a pair of pneumatic artificial muscles in an isometric bench-top configuration at nominal muscle length (27.5 cm). The input signal was a 5-ms square pulse whose amplitude produced a control signal with a 10-volt peak. Vertical lines indicate time points used to calculate electromechanical delay (EMD), time to peak tension (TPT), and half relaxation time (HRT). EMD was the time from the start of the square pulse to the onset of artificial muscle force development (3 standard deviations above baseline force). TPT was the time from the onset of force development to when peak force was achieved. HRT was the time from peak force to when the force dropped 50% from peak force.

Subject Testing

During standing, the subject was able to sustain a maximum voluntary activation while wearing the powered prosthesis. With the artificial plantar flexor muscles fully inflated (control signal of 10 V), the subject was raised up on his toes with his weight distributed between prosthetic and intact sides. While standing wearing the powered prosthesis, the subject was able to alternate between standing with feet flat and standing on his toes without difficulty.

During walking, the subject adapted his residual gastrocnemius muscle activation noticeably within 30 minutes of walking with the powered prosthesis on the treadmill (Figure 3-5). We calculated means and standard deviations of the residual EMG signal using 10 consecutive strides at 1 minute, 5 minutes, and 30 minutes of walking. Data at 1 minute, 5 minutes, and 30 minutes demonstrated

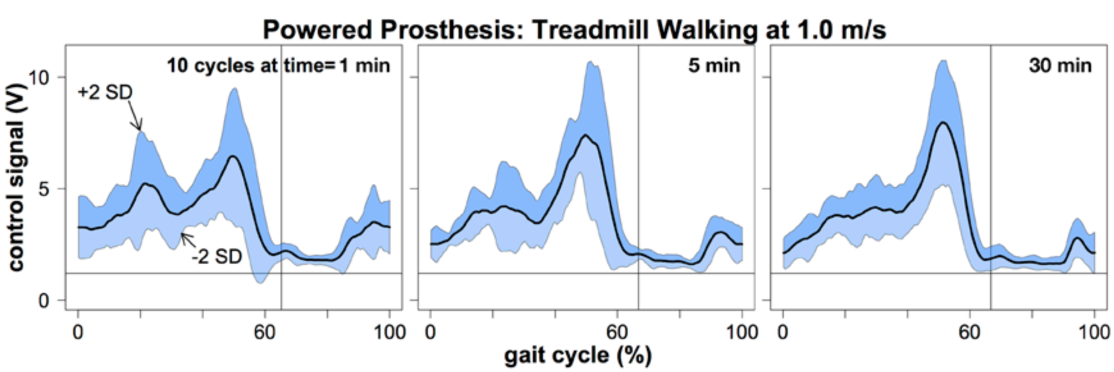

Figure 3-5. Residual Gastrocnemius EMG Control Signal During Walking.
Each plot shows 10 consecutive strides at 1 minute, 5 minutes, and 30 minutes of walking. Black lines show mean of 10 strides. Shaded regions show ±2 standard deviations. Horizontal lines show the 1.2 V offset to achieve the ankle set-point stiffness. Vertical lines show the average toe-off timing of 10 strides.

distinct changes in the subject's muscle activation patterns. When the subject first began to walk with the powered prosthesis, his muscle activity had two high-magnitude peaks during the stance and substantial muscle activity developing at mid swing. After 30 minutes of walking, his muscle activity had one distinct high-magnitude peak in late stance and a small-magnitude peak in late swing. The subject was able to reduce his muscle activity during the swing and maintain a stable low activation. He was also able to produce a more gradual increase in muscle activity through stance.

The subject's ankle angle, moment, and power were similar for his prosthetic and intact sides when he walked with the powered prosthesis (Figure 3-6). After 30 minutes of walking with the powered prosthesis, the subject was able to generate similar peak power during push off between prosthetic and intact sides (Figure 3-6). With his prescribed prosthesis, he generated 3.66 watts/kg of ankle power on his intact side and only 1.40 watts/kg on his prosthetic side at push off. With the powered prosthesis, he generated 3.24 watts/kg of ankle power on his intact side and 3.14 watts/kg on his prosthetic side. The subject's prosthetic ankle range of motion was greater while walking with the powered prosthesis compared to his prescribed prosthesis. The subject's prosthetic ankle range of motion while walking with the powered prosthesis was 30 degrees compared to 18 degrees with his prescribed prosthesis. Compared to walking with his prescribed prosthesis, the subject walked with slightly longer strides on his prosthetic side when wearing the powered prosthesis. The difference in strides lengths can be seen in the toe-off timings as shown in Figure 3-6.

DISCUSSION

We believe that the performance of our experimental powered prosthesis is suitable for walking because the force bandwidth and electromechanical delay of our pneumatic artificial muscles were similar to what has been measured for the human system. The force bandwidth of the pneumatic artificial muscles (2.0 Hz) was similar to that of human skeletal muscle. Dynamic response models of skeletal muscle have reported plantar flexor force bandwidths of 1.6-2.0 Hz [69]. The electromechanical delay of our pneumatic artificial muscles (33 ms) was similar to that of skeletal muscle. The electromechanical delay of plantar flexor skeletal muscle reported in past studies range from 20-30 ms during voluntary contractions [70, 71]. The time to peak tension of our artificial muscles was 48 ms and the half relaxation time was 50 ms. It is difficult to compare our time to peak tension and half relaxation times to in-vivo studies as these contractile properties are significantly different depending on ankle position [72]. For studies where the ankle position was close to full plantar flexion (30 degrees plantar flexion), time to peak tension ranged from 100-125 ms and half relaxation time ranged from 80-125 ms [72, 73]. We could reduce the response time of our system by increasing flow rate (by increasing the number of pressure regulators) or decreasing actuator volume (to reduce dead space) [66]. However, given that we matched the force bandwidth and electromechanical delay of human muscle, we were content with the current design.

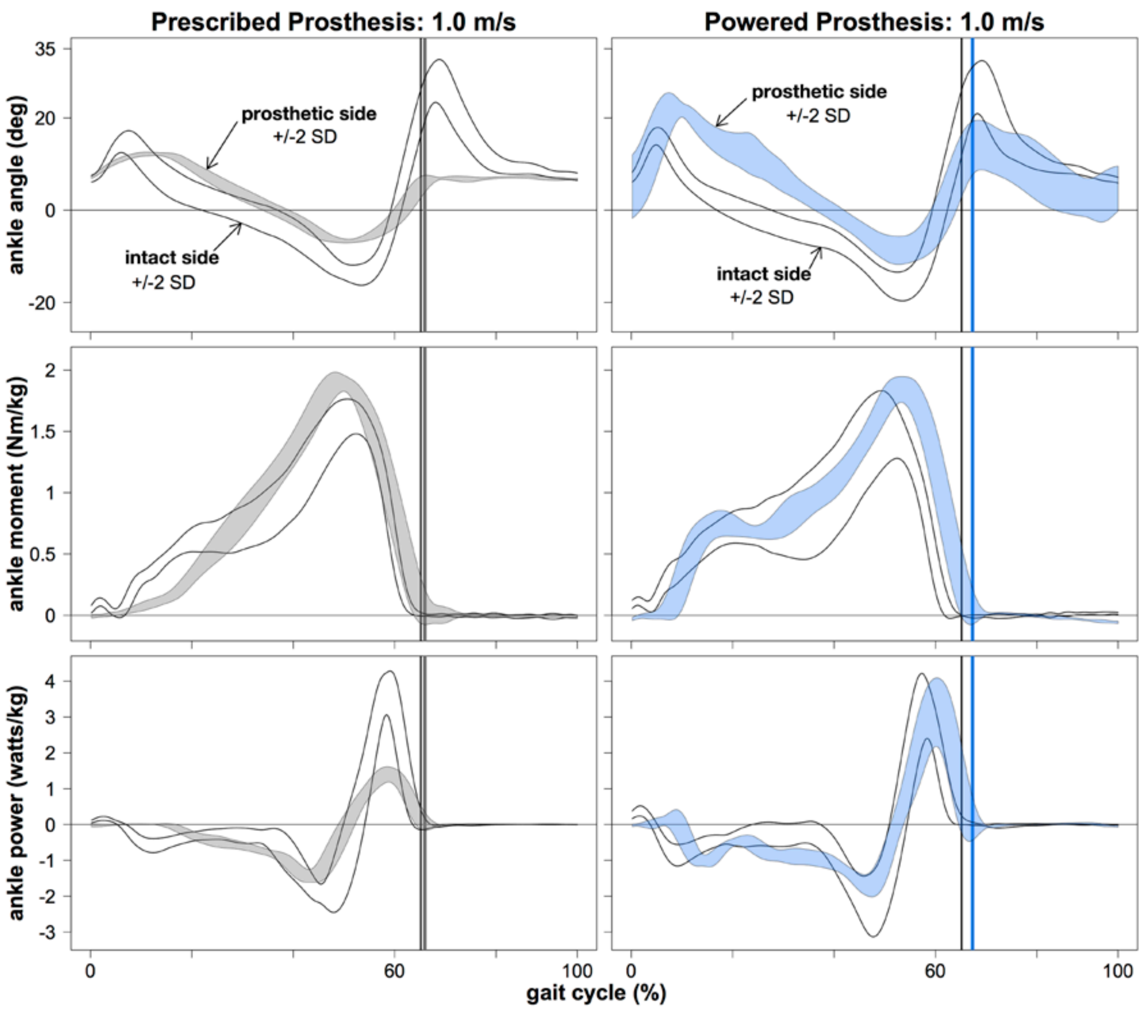

Figure 3-6. Ankle Mechanics of Prescribed vs. Powered Prosthesis.
Ankle angles, moments, and powers were calculated from 10 consecutive cycles. For the powered prosthesis, 10 consecutive cycles starting at the 30-minute time point were used. The proportional EMG control signal for these 10 cycles is shown in Figure 3-5. Outlined regions (intact side) and shaded regions (prosthetic side) show ±2 standard deviations about the mean. Vertical lines show the average toe-off timing for intact and prosthetic sides

We designed the powered prosthesis to serve as a testbed for developing physiologic controllers for powered lower limb prostheses. We can implement proportional myoelectric control of both plantar flexion and dorsiflexion with the powered prosthesis. We can also easily reconfigure the pneumatic artificial plantar flexor and dorsiflexor muscles (i.e. by changing nominal muscle lengths, muscle attachment points, baseline pressures) to achieve a wide range of ankle position, ankle stiffness, and ankle power. In this study, we tuned the pneumatic artificial plantar flexor and dorsiflexor muscles to produce ample plantar flexion torque for push off and provide adequate ankle stiffness for weight acceptance during walking. It should be emphasized that we designed the powered prosthesis solely for in-laboratory experimentation with no intention to develop the device for clinical applications.

The powered prosthesis using proportional myoelectric control of plantar flexion allowed the amputee subject to walk with a more normal gait than his prescribed passive prosthesis. With a short period of practice, the subject modified his residual muscle activation patterns to produce prosthetic ankle power during push off that matched his intact ankle power. He was also able to minimize his residual muscle activation during swing to achieve foot clearance needed to avoid foot scuffing and tripping.

One limitation of our study is that we only tested the powered prosthesis with one amputee subject. Additionally, we only had the subject walk with the powered prosthesis at one treadmill speed. We are uncertain whether other amputees with differing subject characteristics (e.g. age, activity level, time since amputation, reason for amputation, residual muscle volume) would also have the ability to learn to walk with the powered prosthesis to produce symmetric gait. We are also uncertain how learning to walk on a treadmill with proportional myoelectric control would translate to performing other locomotor tasks (e.g. walking over ground, ambulating stairs).

Another limitation of our study is that the subject did not have active control of dorsiflexion when walking with the powered prosthesis. One potential advantage of implementing proportional myoelectric control of dorsiflexion in addition to proportional myoelectric control of plantar flexion during walking is that the amputee could achieve a more desirable roll-over shape. Another potential advantage is that the amputee could control the timing and amount of toe clearance. In this study, we only used one control degree of freedom (i.e. residual gastrocnemius EMG to control plantar flexion) because it was the most direct approach to determine whether utilizing residual muscle activation signals for proportional myoelectric control during walking was even possible. We are uncertain how adding a second control degree of freedom (i.e. residual tibialis anterior EMG to control dorsiflexion) would affect the subject's ability to learn to walk with the powered prosthesis.

CONCLUSION

In this study, we designed and tested an experimental lower limb powered prosthesis that can be used in a laboratory environment to implement proportional myoelectric control during walking. Findings from this study suggest that it is feasible to utilize residual muscle activation signals for proportional myoelectric control during walking as demonstrated by one amputee subject. In our future work, we will recruit more subjects to investigate how amputees with differing subject characteristics learn to walk with the powered prosthesis using proportional myoelectric control to modulate their ankle mechanics throughout gait.

Chapter 4: Locomotor Adaptation by Transtibial Amputees Walking with a Powered Prosthesis Under Continuous Myoelectric Control

This chapter will be summited for peer review:
Journal: *IEEE Transactions on Neural Systems and Rehabilitation Engineering*
Title: "Locomotor Adaptation by Transtibial Amputees Walking with a Powered Prosthesis Under Continuous Myoelectric Control"
Authors: Stephanie Huang, Jeffrey P Wensman, and Daniel P. Ferris

ABSTRACT

Lower limb amputees can use electrical activity from their residual muscles for myoelectric control of a powered prosthesis. The most common approach for myoelectric control is a finite state controller that identifies behavioral states and discrete changes in motor tasks. An alternative approach to state-based myoelectric control is continuous proportional myoelectric control where ongoing electrical activity has a proportional relationship to the prosthetic joint torque or power. To test the potential of continuous proportional myoelectric control for powered lower limb prostheses, we recruited five unilateral transtibial amputees to walk on a treadmill with an experimental powered prosthesis. Subjects walked using the powered prosthesis with and without visual feedback of their control signal in real time. Amputee subjects were able to adapt their

residual muscle activation patterns to alter prosthetic ankle mechanics when we provided visual feedback. During walking with visual feedback, subjects significantly increased their peak prosthetic ankle power and positive work during gait above their prescribed prosthesis values ($p = 0.02$, ANOVA). These results indicate that continuous proportional myoelectric control is a viable approach for powered below-knee prostheses. It provided amputee users with the ability to constantly alter their prosthesis mechanics and may be particularly effective in situations with variable locomotor behaviors or varied terrain.

INTRODUCTION

A powered lower limb prosthesis that uses continuous proportional myoelectric control during walking would increase the adaptability of the prosthesis by giving amputees the ability to freely alter their ankle mechanics. With volitional control of their prosthesis, amputees would be able to perform discrete motor tasks (e.g. standing on toes) and highly variable locomotor tasks (e.g. freestyle dancing) that are difficult or impossible to perform with passive prostheses or finite state controlled robotic prostheses. Volitional control of ankle mechanics would also allow amputees to adapt their ankle to variations in ground compliance or unevenness (e.g. traversing an environment with grassy, muddy, and rock surfaces).

The traditional approach for myoelectric control of powered prostheses is to use finite state controllers [16, 74, 75]. Typical state-based controllers for powered lower limb prostheses use intrinsic sensing to detect transition phases into a finite state, where each finite state is defined by a set of parameters that alter the mechanics (e.g. impedance) of the prosthesis [6, 9, 10, 12, 76, 77]. More recently, some groups have begun integrating proportional myoelectric aspects into traditional myoelectric state control and intrinsic sensor state control [22, 24]. For example, Hugh Herr's group at MIT developed a myoelectric controller within their finite state controller that is active only during the controlled dorsiflexion state. The myoelectric controller computes an estimate of the amputee's residual muscle activation magnitude during controlled dorsiflexion, and the state controller then uses that constant value to scale the ankle torque gain parameter of the powered plantar flexion state [24]. While the feasibility of state-based myoelectric controllers during walking has been demonstrated, there is still potential for more continuous myoelectric controllers to provide expanded options to amputees using powered prostheses.

In a recent study, we demonstrated that one amputee subject was able to adapt his residual muscle to control prosthetic ankle mechanics during walking using continuous proportional myoelectric control [78]. Our results showed that within a short time (i.e. 30 minutes) the amputee was able to generate prosthetic ankle power similar to his intact side. However, we only demonstrated this with a single amputee subject and are uncertain whether other amputees would adapt their residual muscle activation patterns similarly when walking with the powered prosthesis. In addition, the length of training and type of feedback might affect the locomotor adaptation.

The purpose of this study was to determine whether amputees with varying characteristics (e.g. time since amputation, reason for amputation) are able to adapt their residual muscle activation patterns to alter ankle mechanics during walking. We recruited five amputee subjects to walk with an experimental powered prosthesis using their residual muscle for continuous proportional myoelectric control. We asked the subjects to perform a controlled locomotor task of walking on the treadmill at a single speed to examine feasibility. Initially we allowed the subjects to practice walking with only verbal feedback, but later we added visual feedback of the myoelectric control signal to determine if that affected the locomotor adaptation.

METHODS

Experimental Powered Prosthesis

We used an experimental powered prosthesis with pneumatic artificial dorsiflexor and plantar flexor muscles to implement continuous proportional myoelectric control of plantar flexion during treadmill walking [78]. Figure 1 shows an amputee subject walking with the experimental powered prosthesis. We controlled the inflation of the artificial plantar flexor muscles using three proportional pressure controllers (MAC Valves, Wixom, MI) per muscle. We maintained constant pressure in the artificial dorsiflexor muscles using one proportional pressure controller per muscle. The input signal for each pressure controller spanned 0-10 volts and 0-90 psi. We interfaced MATLAB/Simulink with a controller board (dSPACE, Inc., Northville, MI) to compute and send input signals to the pressure controllers in real time. We sent a baseline signal to the plantar flexor pressure controllers (2 V) and constant signal to the dorsiflexor pressure controllers (3.5 V) to achieve set-point ankle stiffness at the neutral (locked) ankle position. The set-point ankle stiffness was the same across subjects.

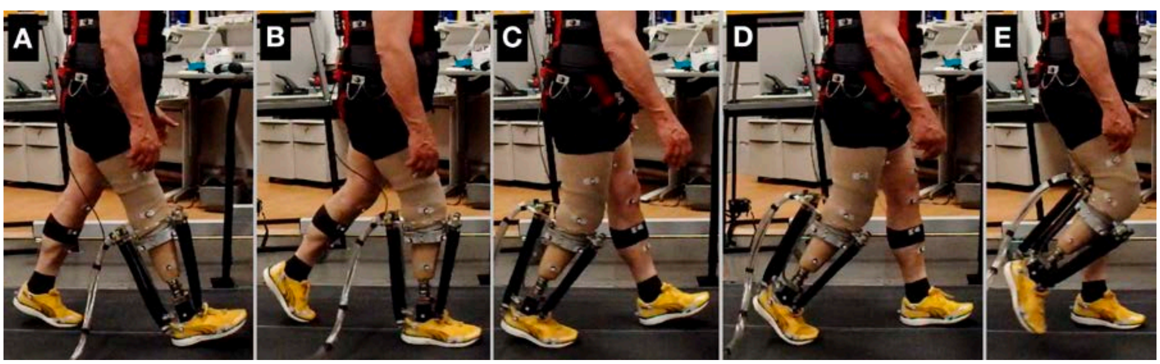

Figure 4-1. Amputee Subject Walking with Experimental Powered Prosthesis.
Experimental prosthesis using pneumatic artificial dorsiflexor and plantar flexor muscles. Amputee subject controls the amount of pressure sent to the pneumatic artificial plantar flexor muscles continuously throughout gait. The control signal is proportional to the subject's smoothed residual gastrocnemius electromyography (EMG) signal. A. Heel Strike, B. Foot Flat, C. Heel Off, D. Toe Off, E. Early Swing. Pressure in the pneumatic artificial plantar flexor muscles is minimum at toe off and maximum at toe off, preceding powered plantar flexion. Maximum dorsiflexion is seen at heel off (B) and maximum plantar flexion is seen at toe off (D).

The inflation pressure of the artificial plantar flexor muscles above baseline was proportional to the residual gastrocnemius electromyography (EMG) signal. To generate a continuous proportional myoelectric control signal we processed the residual EMG in real time. We high-pass filtered the EMG signal (second order Butterworth, high pass, 100 Hz) to attenuate signal artifacts. Then we rectified the high-passed signal (full wave) and low-pass filtered the rectified signal (second order Butterworth, low pass, 4 Hz) to produce a smooth control signal. We applied a subject-specific gain to the smoothed signal in order to scale the output (EMG) range to the input range of the pressure controllers.

Sensors

We recorded surface electromyography (EMG) from either the residual medial or lateral gastrocnemius muscle at 1000 Hz using preamplifier electrodes (Biometrics Ltd, Newport, United Kingdom). To determine EMG sensor placement, we palpated the residual gastrocnemius muscles and chose a recording site over the muscle that was more prominent during a sustained voluntary muscle contraction. We placed the sensor as close to centerline of the muscle belly as possible while avoiding sensitive skin areas and bony protuberances. To minimize skin irritation, we placed silicone putty around the edges of the EMG sensor and between the electrode leads. We also surrounded the EMG sensor with a section of a gel liner with a cut out to reduce signal artifacts generated from movement between the sensor and the skin surface (Figure 4-2).

In addition to surface EMG, we recorded lower-limb segment kinematics at 100 Hz using a Vicon Motion Capture system (Vicon, Lake Forest, CA) and force plate data at 1000 Hz using a split-belt force-measuring treadmill (Bertec, Columbus, OH). Using this data, we were able to calculate ankle angle, ground reaction force, center of pressure, and ankle moment [68]. For the prosthetic side, we defined the ankle joint center from a motion capture marker placed at the prosthetic ankle center of rotation. For the intact side, we estimated the ankle joint center from a motion capture marker place on the lateral malleolus.

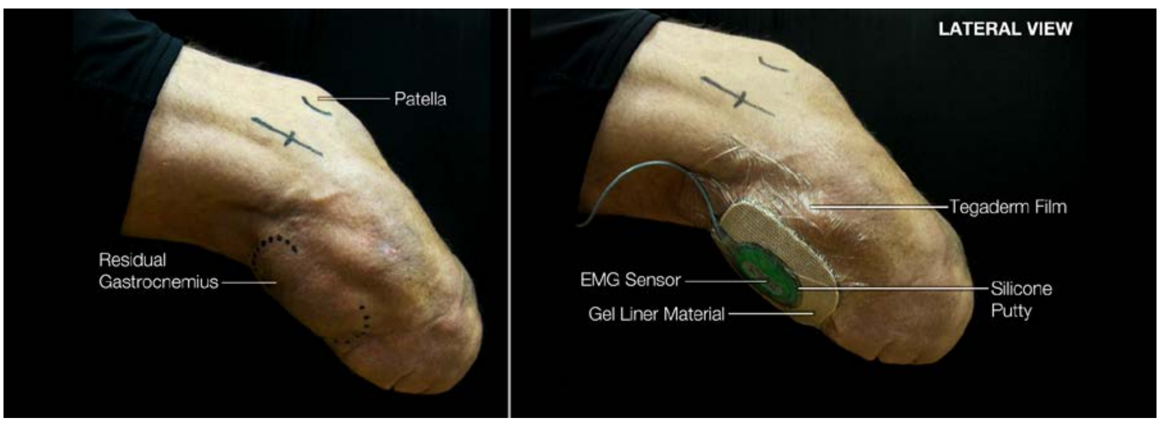

Figure 4-2. EMG Sensor Placement on Residual Gastrocnemius Muscle.
Approximate position of residual gastrocnemius muscle determined via palpation during voluntary muscle contraction. Silicone putty placed around the outside edge of the sensor and around the beginning section of the lead wire. Gel liner material with cut out for sensor butted up against the silicone putty. Tegaderm film placed over sensor and gel liner. Tegaderm was not used for subjects susceptible to skin irritation or breakage.

Subject Testing

We recruited five unilateral transtibial amputees to walk on a treadmill while wearing the experimental powered prosthesis using continuous proportional myoelectric control [78]. Subjects provided informed written consent to a protocol previously approved by the University of Michigan Institutional Review

Board for the Protection of Human Subjects. Subject characteristics are listed in Table 1. Subjects wore their prescribed prosthetic liner and socket with the experimental powered prosthesis. A certified prosthetist from the University of Michigan Orthotics and Prosthetics Center determined proper socket alignment for each subject. We asked subjects to walk on the treadmill at a constant speed of 1.0 m/s using the powered prosthesis. Before walking with the powered prosthesis, we had subjects walk on the treadmill with their prescribed prosthesis until they felt confident walking on the treadmill. We required subjects to wear a body weight support harness (providing no active support) while walking to prevent falls. The harness also reduced the subject's level of stress associated with the risk of falling while walking with the device.

Table 4-1. Amputee Subject Details

Subject	Reason for Amputation	Age (years)	Post-Amputation (years)	Muscle Fatigue * (level)
A	Trauma (Motorcycle Crash)	65	44	3
B	Cancer (Osteogenic Sarcoma)	23	12	3
C	Pain (Nonunion Ankle)	70	10	2
D	Electrical Burn (Electrocution)	60	43	1
E	Trauma Burn (Plane Crash)	59	4	1

* Muscle fatigue level determined from maximum walking ability per trial with the powered prosthesis. Level 1 = 10 or more minutes, level 2= 5-10 minutes, level 3= less than 5 minutes.

We set an initial controller gain for each subject at the beginning of each testing session. At the initial gain level, subjects were able to stand on their toes to achieve maximum ankle plantar flexion with submaximal muscle activation. Between walking trials, we allowed the subjects to increase or decrease the controller gain. It was helpful for some subjects to decrease the gain and increase the gain gradually as they started adapting their residual muscle activation. This was often the case for subjects who experienced repetitive involuntary or unintended residual muscle activation patterns that generated perturbations that they were not able to recover from in subsequent strides (i.e. subject not able to regain control of their residual muscle activation to the point where we needed to stop the treadmill because they were unable to take steps).

All subjects participated in two testing blocks separated by six months. Each testing block spanned one month. In the first testing block, subjects walked with the powered prosthesis with no visual feedback display. In the second testing block, we provided subjects with a visual feedback display of their control signal in real time. Table 4-2 provides a breakdown of the amount of time each subject walked with the powered prosthesis. Subjects participated in one testing session per testing day and completed as many walking trials as they were able to complete during the session. We ended each walking trial when the subject indicated they wanted to stop or we found it necessary to stop. We ended a testing session early if the subject had discomfort that we could not resolve or if the subject could not recover from residual muscle fatigue despite resting periods between walking trials.

During the first testing block, we verbally instructed the subjects on how to activate their residual muscle to control the powered prosthesis. We told them to keep their residual muscle relaxed throughout swing and to gradually ramp up their residual muscle activation throughout stance. We also indicated that their peak muscle activation should occur towards the end of stance. Between walking trials, we reemphasized these general instructions. We offered additional verbal instruction if we thought it would help the subject understand how to control the prosthesis better or if the subject asked for guidance.

During the second testing block, we provided subjects with visual feedback of their residual muscle control signal and a target activation profile in real time. The visual feedback was displayed on a computer monitor fixed to the body weight support system at a height where the subject felt comfortable looking at the display while walking. Figure 4-3 shows the visual feedback display in detail. We instructed subjects to try and generate a control signal within the target area shown on the display. We told them to follow the general shape of the target area, but that the magnitude and timing of the control signal peak would be different for each person. We encouraged the subjects to explore the visual feedback space and get a sense for how the shape of their control signal affected the behavior of the powered prosthesis. Before walking with the visual feedback using the powered prosthesis, subjects walked with visual feedback using their prescribed prosthesis to get used to the information provided.

Table 4-2. Total Time Walking with Powered Prosthesis By Subject

Testing Block #1 – No Visual Feedback Display

Subject	Sessions (count)	Trials/Session (mean)	Minutes/Trial (mean)	Total Minutes (sum)	Cumulative Hours
A	5	6	4	146	2.4
B	4	9	5	182	3.0
C	2	10	4	79	1.3
D	4	5	6	95	1.6
E	4	6	7	158	2.6
mean(SD)	4(1)	7(2)	5(1)	132(43)	2.2(0.7)

Testing Block #2 – With Visual Feedback Display

Subject	Sessions (count)	Trials/Session (mean)	Minutes/Trial (mean)	Total Minutes (sum)	Cumulative Hours
A	3	6	4	66	3.5
B	3	6	4	54	3.9
C	3	6	4	77	2.6
D	5	4	5	94	3.2
E	2	8	7	101	4.3
mean(SD)	3(1)	6(1)	5(1)	78(19)	4(1)

Biomechanics Analysis

For each subject we analyzed data from their final trial on the last day of each testing block. We selected the segment of 20 consecutive strides where the subject generated the most similar control signal across strides. To select the 20 consecutive strides, we segmented the trial using a 20-stride sliding window with a 15-stride overlap and calculated the mean control signal. For each segment, we calculated the cross correlation at zero lag [34] for each stride against the 20-stride mean and selected the trial segment with the highest mean (of 20) cross correlation values. This was necessary because muscle fatigue and motor exploration by the subjects both led to relatively high variability across and within the training sessions.

We compared residual limb muscle activation patterns, ankle power, and ankle work across walking conditions (prescribed prosthesis, powered prosthesis with no visual feedback, powered prosthesis with visual feedback). We estimated ankle power as the vector dot product between ankle moment and ankle angular velocity. We estimated external ankle work as the time-integral of the ankle power curve. We analyzed the positive work component to quantify the amount of energy generated by the powered prosthesis for forward propulsion. We also analyzed negative work during early stance for the powered prosthesis

Statistical Analysis

We performed repeated measures ANOVAs to determine if there were significant differences in peak ankle power, total ankle work, positive ankle work, and negative ankle work between walking condition and subject. For all ANOVAs, if walking condition or subject was a significant factor ($p<0.05$), we performed a Tukey's Honestly Significant Difference test to determine which contrasts were significant ($p<0.05$).

Figure 4-3. Subject Walking with Visual Feedback Displayed on Monitor.
Shaded area is the target region. Subject is instructed to generate control signal anywhere within the target region that follows the approximate shape of the target region. Control signal is drawn in real time. Vertical line is the Subject's approximate toe off timing. At toe off, an open circle point character is drawn at the control signal value. The subject is told that the open circle point character should appear slightly before, on, or slightly after the vertical line. Line widths are thick during stance and lighter during swing, determined by vertical ground reaction force measured from the prosthetic-side force plate. The x-axis (time) and y-axis (control signal) limits were fixed. Stance phase time (start of x-axis to vertical line) and swing phase time (vertical line to end of x-axis) were estimated for each subject from the average heel strike and toe off timing calculated while walking with their prescribed prosthesis on the treadmill at 1.0 m/s. The visual display foreground drew the control signal in real time. When the foot was on the ground, the control signal and vertical toe-off line was drawn with a thick line width. When the foot was off the ground, the control signal and vertical toe-off line was drawn with a thin line width. At the instant of toe off, an open circle point character was drawn at the time and control signal coordinates. At the instant of heel strike, the foreground was cleared and the control signal was replotted. We used vertical ground reaction force to detect when the foot heel strike and toe off events.

RESULTS

Subjects adapted their residual muscle activation patterns to increase ankle power when we provided them with visual feedback of their control signal (Figure 4-4, Figure 4-5). There was a significant increase in peak ankle power ($p=0.02$) when subjects walked with the powered prosthesis with visual feedback when compared to walking with their prescribed prosthesis (Table 4-3). Subjects' control signals appeared similar in shape, but the timing and magnitude of their control signal peaks relative to toe off were variable (Figure 4-6). Consequently, we saw large differences in the magnitude of peak ankle power and variation in the timing of peak power relative to toe off (Figure 4-4, Figure 4-5).

Without visual feedback, subjects did not adapt their residual muscle activation patterns to increase ankle power during walking (Figure 4-4, Figure 4-5). Four of five subjects did not walk with increased ankle power over their prescribed prosthesis (Figure 4-4). There was no significant difference in peak ankle power compared to when subjects walked with their prescribed prostheses (Table 4-3). Without visual feedback, subjects' control signals did not converge to a similar shape as seen when visual feedback was provided to the subjects. Two subjects exhibited sustained activation through mid stance, which limited their capacity to generate ankle power (Figure 4-4). Two subjects had low activation at heel strike that gradually increased through stance, but their peak activation was not high enough to generate ankle power beyond that of their prescribed prosthesis. One subject (Subject A) produced a control signal with a distinct, high amplitude peak during late stance, which generated substantial ankle power. This subject also had an undesirable high amplitude, short duration control signal peak following heel strike as a result of involuntary residual muscle activations.

Subjects increased their total ankle work while walking with the powered prosthesis when we provided them with a visual feedback display of their control signal and all but one subject (Subject E) produced net positive ankle work (Figure 4-6). There was a significant increase in total ankle work (vs. prescribed, $p=0.02$; vs. no visual feedback, $p=0.04$) and positive ankle work (vs. prescribed, $p= 0.02$) when subjects walked with the powered prosthesis with visual feedback when compared to walking with their prescribed prosthesis and the powered prosthesis without visual feedback (Table 4-3). There was no significant difference in negative ankle work (Table 4-3). Although all subjects increased their ankle work with visual feedback, they walked with vastly different gait patterns. Differences in gait patterns across subjects are clear to see when looking at each subject's characteristic ankle work loop (Figure 4-6). The subjects' control signals that correspond to ankle work loops are also shown in Figure 4-6. Three noticeable differences between subjects' control signals that align with differences seen in their gait patterns are: 1) the magnitude of the control signal at toe off, 2) the rate of change of the control signal from heel strike to peak activation, and 3) the rate of change of the control signal from peak activation to toe off.

When subjects were not provided the visual feedback display, their ankle work remained similar to when walking with their prescribed prostheses (Figure 4-6). Only one subject showed an increase ankle work that resulted in net positive ankle work (Subject A). There was no significant difference in total ankle work, positive ankle work, or negative work when the subjects walked without visual feedback compared to walking with their prescribed prosthesis (Table 4-3).

DISCUSSION

The main finding of this study was that the amputee subjects were able to adapt their residual muscle activation patterns during walking to continuously control their prosthetic ankle mechanics using proportional myoelectric control. Without visual feedback, subjects did not converge to a similar residual muscle activation pattern, but they were able to walk with the powered prosthesis and generate similar ankle mechanics to their prescribed prosthesis. With visual feedback, subjects were able to produce a control signal that resembled the general target shape shown on the visual display. Not only were the subjects able to converge to a similar residual muscle activation pattern, but also were able to generate greater ankle power and external ankle work compared to their

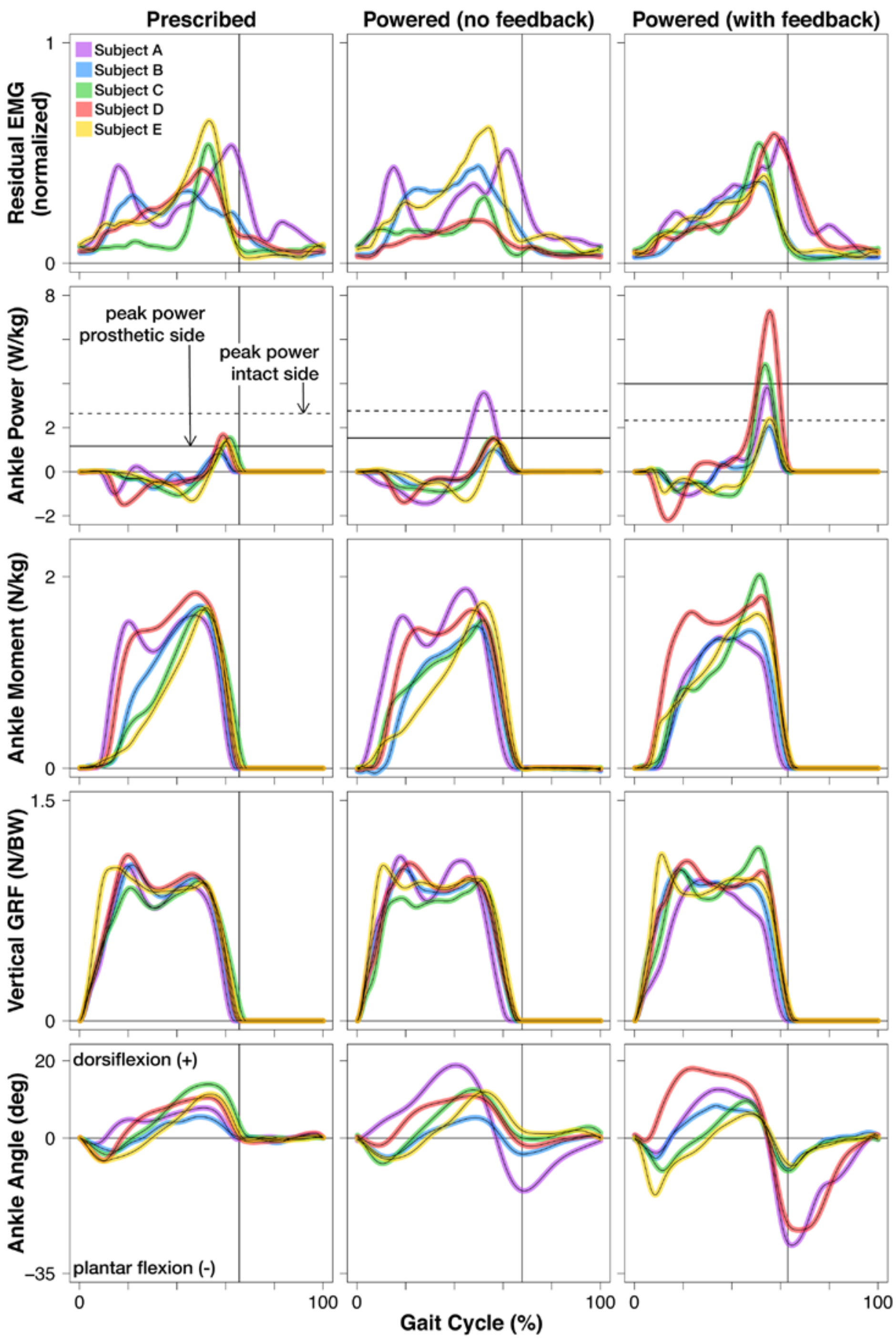

Figure 4-4. Residual EMG Patterns and Ankle Mechanics.
Residual EMG Patterns and Ankle Mechanics. Vertical line shows group average toe off. Each curve represents an individual subject and shows the subject's mean of 20-consecutive cycles. With visual feedback, all subjects walked with increased prosthetic ankle power. Variability in peak ankle power with visual feedback was high compared to variability in peak ankle power seen when subjects walked using their prescribed prosthesis

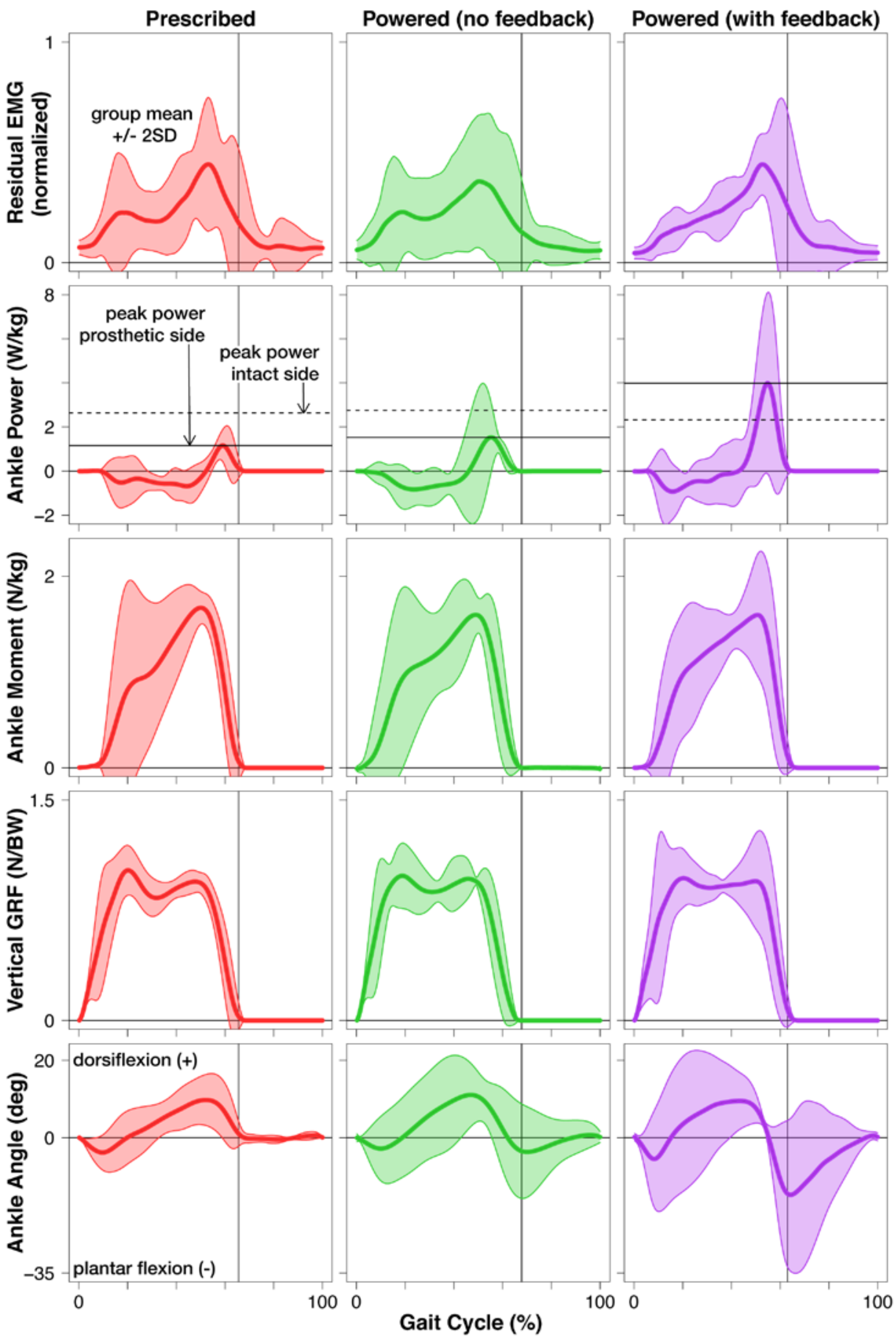

Figure 4-5. Residual EMG Patterns and Ankle Mechanics: Group Mean
Residual EMG Patterns and Ankle Mechanics: Group Mean. Shaded area is ±2 standard deviations about group mean. Group mean calculated from subject mean of 20 consecutive cycles. Vertical line shows group average toe off. On average, subjects did not choose to walk with the powered prosthesis to increase prosthetic ankle power when they were not provided visual feedback of their control signal. With visual feedback, all subjects walked with increased prosthetic ankle power. Variability in residual muscle activation patterns was substantially smaller with feedback compared to without. Variability in ankle angle was substantially greater with feedback compared to without.

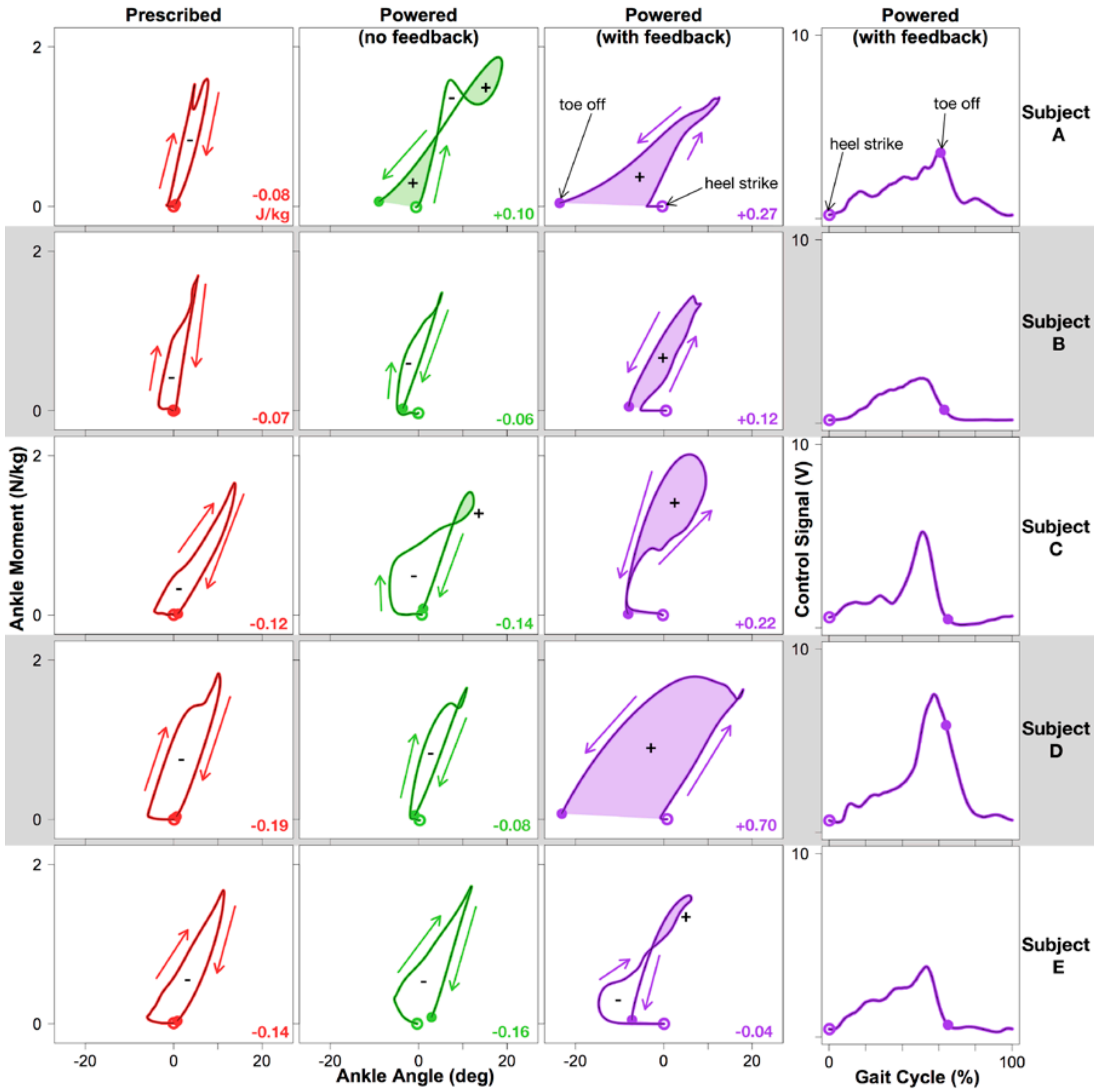

Figure 4-6. Ankle Work Loops and Control Signal.
Ankle Work Loops and Control Signal. Ankle angle vs. ankle moment for five subjects (Subjects A-E). Curves show mean of 20-consecutive cycles. Counter-clockwise loops indicate positive work; clockwise loops indicate negative work. The right-most column shows each subject's corresponding control signal when walking with visual feedback across the same 20 strides shown for their work loops. Net external ankle work values are given in the lower right corner of the work loops plots. All subjects increased net external ankle work during walking with the powered prosthesis with visual feedback. All but one subject had net positive ankle work when walking with visual feedback. Gait patterns of subjects walking with the powered prosthesis with visual feedback were extremely variable as seen from the stark differences in work loop shapes.

Table 4-3. Prosthetic Ankle Peak Power and Ankle Work

Prosthetic Ankle Peak Power [W/kg]

Subject	Prescribed* mean (sd)	Powered- no feedback° mean (sd)	Powered- with feedback*° mean (sd)
A01	1.098 (0.027)	3.688 (0.900)	4.830 (1.907)
A02	0.839 (0.053)	1.037 (0.439)	2.210 (0.710)
A03	1.550 (0.068)	1.565 (0.569)	5.075 (1.653)
A04	1.701 (0.117)	1.512 (0.258)	8.144 (1.621)
A05	1.376 (0.084)	1.318 (0.287)	2.648 (1.388)

Prosthetic Ankle Net Work [J/kg]

Subject	Prescribed* mean (sd)	Powered- no feedback° mean (sd)	Powered- with feedback*° mean (sd)
A01	-0.075 (0.004)	0.102 (0.088)	0.274 (0.160)
A02	-0.073 (0.005)	-0.057 (0.061)	0.124 (0.109)
A03	-0.115 (0.006)	-0.135 (0.083)	0.215 (0.174)
A04	-0.193 (0.007)	-0.080 (0.036)	0.698 (0.106)
A05	-0.143 (0.003)	-0.156 (0.046)	-0.035 (0.158)

Prosthetic Ankle Positive Work [J/kg]

Subject	Prescribed* mean (sd)	Powered- no feedback mean (sd)	Powered- with feedback* mean (sd)
A01	0.093 (0.005)	0.487 (0.094)	0.551 (0.161)
A02	0.081 (0.004)	0.121 (0.051)	0.292 (0.089)
A03	0.149 (0.004)	0.198 (0.070)	0.512 (0.168)
A04	0.130 (0.006)	0.180 (0.036)	1.002 (0.143)
A05	0.117 (0.007)	0.152 (0.033)	0.288 (0.157)

Prosthetic Ankle Negative Work [J/kg]

Subject	Prescribed mean (sd)	Powered- no feedback mean (sd)	Powered- with feedback mean (sd)
A01	-0.168 (0.006)	-0.386 (0.044)	-0.277 (0.063)
A02	-0.154 (0.006)	-0.179 (0.038)	-0.169 (0.040)
A03	-0.264 (0.008)	-0.333 (0.040)	-0.297 (0.020)
A04	-0.324 (0.006)	-0.260 (0.023)	-0.303 (0.053)
A05	- 0.260 (0.007)	-0.308 (0.037)	-0.324 (0.055)

*$p<0.05$, Powered Prosthesis- with feedback vs. Prescribed
°$p<0.05$, Powered Prosthesis- with feedback vs. Powered Prosthesis- no feedback

prescribed prosthesis. The fact that the subjects were able to substantially change their residual muscle activation patterns when we added visual feedback suggests that amputees may need explicit guidance on potential control strategies when using continuous proportional myoelectric control.

Although the subjects' residual muscle control signals were similar to each other with visual feedback, their resulting gait patterns were very different from each other. Small differences in timing, magnitude, and shape of the control signal resulted in large differences in the ankle mechanics of the powered prosthesis during walking. Subjects' choice of body posture, intact joint kinematics, and limb loading also contributed to the differences seen across their gait patterns. The force in the artificial pneumatic muscles used for the prosthesis has length-dependent properties that will influence the power output [66, 67, 79]. In addition, the rate of change in activation signal affects power output due to the activation dynamics of the pneumatic muscles. Figure 6 reveals differences in the slope of rising activation during the first half of the stance phase and the timing of muscle relaxation. Subject D's rate of increase in muscle activation to peak activation and his delayed relaxation resulted in large positive ankle work. In fact, the ankle prosthesis had greater mechanical work than what is normally observed in walking by intact subjects [80].

There are several possibilities why the amputees did not adapt their muscle activation patterns to increase peak ankle power when they were not provided visual feedback. The two most likely possibilities were: 1) the subjects were lacking feedback necessary to understand how the powered prosthesis was behaving, and 2) the subjects chose not to walk with the powered prosthesis to alter ankle mechanics despite their capability to do so. During the no visual feedback condition, we did not explicitly tell the subjects that their goal was to walk with greater ankle power than their prescribed prosthesis. However, all of the subjects were aware that the powered prosthesis was capable of outputting high ankle torque and power, which they experienced when they controlled the powered prosthesis to support their body weight on their prosthetic side. The two most confident subjects (Subjects D and E) later informed us that their strategy when walking with the powered prosthesis without visual feedback was to walk with the "least amount of energy possible".

Providing subjects with visual feedback of their control signal in real time established a more explicit link between subject's neural commands and prosthetic ankle mechanics. Some subjects indicated that the visual feedback convinced them that they were in direct control of their ankle. Through better understanding of the myoelectric controller, the subjects may have been more likely to integrate the dynamics of the prosthetic-controller system into their motor planning because they more fully explored the motor state space [81-83].

The direct visual feedback provided reinforcement of the biomechanical consequences of the neural signals they produced. Subjects could sense step-to-step variability and correlate it with the command signal they generated. It is reasonable to believe that with a longer amount of time walking with the powered prosthesis, the amputees may be able to use non-visual sources of feedback (e.g. angular velocity of the knee, pressure distribution within the socket) for greater locomotor adaptation. Future studies would need to use fully portable powered prostheses that subjects could wear for long durations in their everyday lives. There are several prostheses that could be adapted to continuous proportional myoelectric control [10, 84-86].

There were several limitations to our study. The biggest limitation was that the subjects walked with the powered prosthesis for a relatively short time. As mentioned above, a portable prosthesis could allow subjects to have additional time for practice and locomotor adaptation. Another factor that was somewhat surprising was the amount of fatigue experienced by the users. Some of the subjects quickly became fatigued in their residual limb muscles, as they had not been actively controlling those muscles since amputation. The comfort of the socket interface was also an issue. A powered prosthesis increases shear forces and interface torques compared to a passive prosthesis due to energy transfer from the device to the user. We believe that with longer exposure time to walking with the experimental prosthesis that the amputee subjects would have continued adapt their residual muscle activation patterns as they become more confident walking with the device.

Another limitation of the study is that we used surface electrodes, which are susceptible to signal noise from motion artifacts. While the limb-socket dynamics were relatively stable with subjects' prescribed prostheses, we learned that the limb socket dynamics were much more challenging with our experimental powered prosthesis. In a previous study, we demonstrated that it was possible to record artifact-free signals from residual muscles of transtibial amputees at the limb-socket interface during walking. However we believe that the dynamics of the powered prosthesis translated to micro-motions at the limb-sensor-socket interface, which resulted in signal noise with frequencies that were similar to physiologic muscle activation signals. This was problematic because it was challenging to filter out noise components in the signal. In order to minimize signal noise, we had to modify the limb-sensor-socket interface from our previous study (Figure 1). Our continuous proportional myoelectric controller might have presented changes in pressure distribution at the limb-socket interface that were much more volatile than with a passive device. Also, the dynamics between the ground and our powered prosthesis might have increased vibration at the limb-socket interface. The increase in socket disturbance (i.e. pressure, vibration) likely exacerbated signal noise. In the future, intramuscular electromyography sensors should be able to greatly attenuate or even eliminate the sensor noise issue [30, 87-89]. These sensors are now being used in human experiments and hold the potential to drastically alter the quality and reliability of myoelectric control for powered prosthetic limbs.

CONCLUSION

In this study, we demonstrated that it was possible to implement continuous proportional myoelectric control for an experimental powered lower limb prosthesis during walking. There were challenges with obtaining a robust and reliable electromyography signal from within the limb-socket interface but it is possible to overcome these challenges in short training periods. The results suggest that continuous proportional myoelectric control allows the user to adapt their ongoing prosthesis mechanics to accommodate the motor task.

Chapter 5: Lower Limb Amputee User Experience Walking with a Transtibial Powered Prosthesis Using Continuous Proportional Myoelectric Control

This chapter will be summited for peer review:
Journal: *Disability & Rehabilitation: Assistive Technology*
Title: "Lower Limb Amputee User Experience Walking with a Transtibial Powered Prosthesis Using Continuous Proportional Myoelectric Control"
Authors: Stephanie Huang, Jeffrey P Wensman, and Daniel P. Ferris

ABSTRACT

In a previous study, we demonstrated that amputees were able to walk with an experimental powered prosthesis that uses residual muscles for continuous proportional myoelectric control to dictate ankle torque. Compared to passive prostheses and powered prostheses that use state-based control, a powered prosthesis that uses pure volitional control may have a stronger affect on a user's prosthetic embodiment. The purpose of this study was examine the user experience of amputees walking with a powered prosthesis that uses continuous proportional myoelectric control to gain a better understanding of how the amputee interacts with the prosthesis not only functionally, but also emotionally. We conducted post-study phone interviews for the five amputee subjects that walked with our experimental powered prosthesis in our previous

study. We analyzed phone interview responses along with verbal feedback compiled throughout the study by extracting core themes from the subjects' responses and examining specific subject responses under each core theme. We also composed subject narratives based on first hand observations. The results of our study showed that the amputees had vastly different experiences walking with the powered prosthesis that could be described across seven core themes: *User Device Interface*, *Mental Factors*, *Physical Factors*, *Learning Mechanisms*, *Human Performance*, *Significance*, and *Embodiment*. Our study confirms that one of the biggest challenges to myoelectric control in the lower limb is designing a robust limb-sensor-socket interface. One of the most compelling results of the study was that most subjects who walked with our experimental powered prosthesis experienced some degree of prosthetic embodiment, especially with regards their "sense of self". This study suggests that a powered prosthesis that uses continuous proportional myoelectric control has the potential to provide amputees with increased prosthetic functionality along with substantial emotional benefits.

INTRODUCTION

Powered lower limb prostheses are beginning to incorporate direct (i.e. volitional) control into their control architecture [19-24, 90, 91]. A subset of these devices use electromyography signals recorded from residual muscles for volitional control within a finite state controller. In a state-based myoelectric controller, the embedded myoelectric controller calculates an estimate of the amputee’s residual muscle activation level during one state and outputs the discrete value to scale joint parameters in a second state [22, 24]. It is also possible to implement a myoelectric controller outside the confines of a finite state machine. In a recent study, we used an experimental powered prosthesis to implement a continuous proportional myoelectric controller during walking, where ankle joint torque was directly controlled by the amputee’s smoothed residual muscle activation signal throughout gait (see Chapter 4).

In addition to increasing physical functionality during walking, a powered lower limb prosthesis that uses the amputee's nervous system to control the behavior of their prosthesis has the potential to provide psychological benefits [92-95]. Prosthetic embodiment, generally defined as how much a prosthesis feels like “part of you”, has both functional and emotional components. A prosthesis can feel like it is more “part of the user” functionally by enabling them to perform a task and emotionally by providing a stronger “sense of self” [92]. In the upper limb, myoelectric control of powered prostheses has been shown to extend phantom sensations [92] and reduce the severity of phantom limb pain [37]. In the lower limb, myoelectric control of powered prostheses has not been as widely studied. As the development of direct myoelectric control of powered lower limb prostheses progresses, it will become increasingly important to examine how users interact both physically and emotionally with the prosthesis in order to gain insight for future controller and interface design.

In our previous study, five unilateral transtibial amputee subjects demonstrated that they were able to walk with an experimental powered prosthesis using their residual muscles for continuous proportional myoelectric control (see Chapter 4). We analyzed ankle biomechanics and found that subjects were able to use the experimental powered prosthesis to increase their ankle power and ankle work above that of their prescribed prosthesis. Although all of the subjects were able to walk with the powered prosthesis to directly control their ankle mechanics, we noticed that the subjects' emotional responses to walking with the experimental prosthesis did not necessarily align with what we'd expect from their quantitative results.

The purpose of this study was to examine the user experience of walking with a powered prosthesis using continuous proportional myoelectric control. We collected subjective response data from all five subjects that walked with our experimental powered prosthesis in our previous study. We analyzed post-study interview responses in addition to verbal feedback that we recorded throughout the course of the study.

METHODS

Subjects

We conducted this user experience study with the five unilateral transtibial amputee subjects who took part in our previous study where we asked subjects to walk with an experimental powered prosthesis using continuous proportional myoelectric control from their residual muscles (see Chapter 4). Subject characteristics are shown in Table 5-1.

Table 5-1. Amputee Subject Details

Subject	Reason for Amputation	Age (years)	Post-Amputation (years)	Fatigue * (level)	Confidence ** (rank)
A	Trauma	65	44	3	2
B	Cancer	23	12	3	1
C	Pain	70	10	2	3
D	Electrical Burn	60	43	1	4
E	Trauma Burn	59	4	1	5

* Muscle fatigue level determined from maximum walking ability per trial with the powered prosthesis. Level 1 = 10 or more minutes, level 2= 5-10 minutes, level 3= less than 5 minutes.
** Confidence level when walking with experimental powered prosthesis (ranked via direct observations of lead researcher). Rank 1= least confident, Rank 5= most confident.

Interviews

We conducted two phone interviews with the amputee subjects consisting of a series of open-ended and semi-structured questions (Table 5-2). We conducted the first interview after the subject's completion of the first testing block (without visual feedback) and the second upon the subject's completion of the second testing block (with visual feedback). The goal of the interviews was to generate a casual conversation with each subject in order to gain perspective on their experience participating in the study and walking with the experimental powered prosthesis. We instructed the subjects to answer each question freely and to talk about anything that came to mind that was relevant to the study, even if it was not directly related to the question posed. We allowed the subject to talk freely until they reached a sustained pause. If we felt that the subject was misinterpreting the question, we allowed the subject to complete their response, then asked open-ended probes to guide them towards our intended interpretation of the question. All phone interviews were conducted within one month of the subject's final testing sessions. On average, phone interviews lasted one hour.

Verbal Feedback

In addition to phone interviews, we also compiled verbal feedback from the subjects that we recorded between walking trials with the prosthesis. During the final testing sessions of both testing blocks, we prompted subjects for feedback between walking trials. Examples of questions that we asked subjects are: "How did you feel walking that trial?", "What were your strategies for that trial?", "I noticed that you were landing with your foot flat instead of with a heel strike. Were you doing that deliberately?", "I noticed that your muscle activation was a bit jittery that trial. Does your residual muscle feel tired?". The purpose of asking subjects for feedback between walking trials was to get an idea of the subjects' thought processes, strategies, self-awareness, and self-assessment after walking with the powered prosthesis.

Core Themes

We transcribed, verbatim, the phone interviews and verbal feedback from the final testing sessions. The semi-structured questions that we included in the analysis are listed in Table 5-2. After reading the transcripts and highlighting key sections, we identified core themes that emerged from the subjects' responses. We parsed each subject's interview and verbal feedback transcripts according these core themes. We compiled all of the subjects' data and sorted the parsed data by theme, then grouped similar responses to derive take-home messages from each theme.

Narratives

We composed narratives for each subject based on our first hand observations throughout the course of the entire study. The purpose of the narratives is to try and provide a description of each subject's character. We thought this was important because we noticed that each subject's personality had some effect on how they approached the subject and their general attitude towards the experiment (e.g. excited, indifferent).

Word Frequencies

We generated word frequency visualizations from each subject's responses in the first phone interview (no visual feedback) using TagCrowd.com. We chose to analyze the contents of the first phone interview versus the second (with visual feedback) because the first interview was more representative of each subject's initial experience walking with the device, which is what we wanted to capture. We defined the minimum frequency count to be 5 and the maximum number of words to be 30. We did not include words in the frequencies counts that had no apparent meaning (i.e. “yes”, “no”, “maybe”, “sometimes”, “thing”, “really”, “sure”, “lot”, “going”).

Table 5-2. Phone Interview Questions

1. What thoughts or feelings did you have when you first sat down and were able to move the ankle using your residual muscles?
2. What thoughts or feelings did you have when you first were able to stand on your toes using the powered prosthesis?
3. Can you talk briefly about your overall experience walking with the experimental prosthesis?
4. Right now, you walk with a passive prosthesis that you cannot control. Do you think it is important for you to be able to control your prosthesis?
5. Imagine one day there is a commercially available prosthesis that you could control directly with a similar concept as what you walked with in lab. Do you think this would affect your quality of life? Do you think it would change the way others perceive you or the amputee community as a whole?
6. Did you have any fears or concerns at any time during the experiment? If yes, do you think it affected the way you behaved during testing?
7. Following testing sessions after you left the lab, did you find yourself thinking about anything?
8. Before testing sessions as you were on your way to the lab, did you find yourself thinking about anything?
9. Did you feel physically fatigued during any of the testing sessions?
10. Did you feel mentally fatigued during any of the testing sessions? If yes, did you find that it affected your ability to activate your residual muscles?
11. How successful did you feel walking with the experimental prosthesis? What would have helped you be more successful?
12. Do you think that the experiment you participated in is valuable?
13. Is there anything else you would like to add? Do you have any general comments or thoughts that you would like to tell me, or anything specific that you think is important for me to know moving forward?

RESULTS

Seven core themes emerged from the subjects’ interviews and verbal feedback: *Device User Interface*, *Mental Attributes*, *Physical Attributes*, *Learning Mechanisms*, *Human Performance*, *Significance*, and *Embodiment*. *Device User*

Interface included any element of the residual limb muscle-sensor-prosthetic socket system. *Mental Attributes* included anything related to the subjects' mental fatigue, mindset, or cognitive processes. *Physical Attributes* included anything related to the subjects' body. *Learning Mechanisms* included anything related to how the subjects learned or what might help in the learning process. *Human Performance* included anything related to the subjects' ability to control their residual muscles and/or walk with the powered prosthesis. *Significance* included comments on the importance of the research and technology on the lives of amputees. *Embodiment* comments reflect how users felt about adopting the prosthesis as a part of their body.

We divided the core themes into two groups. The first group encompassed themes relevant to how the amputee interacted with the experimental prosthesis (i.e. *Device User Interface*, *Mental Attributes*, *Physical Attributes*, *Learning Mechanisms*, and *Human Performance*). Themes in this first group were inter-related with each other (Figure 5-1) and reflected the user's experience with the prosthesis while they used it. The second group of themes included *Significance* and *Embodiment*. These two core themes included the user's reflections on using the prosthesis more so than their actual experience while using the prosthesis.

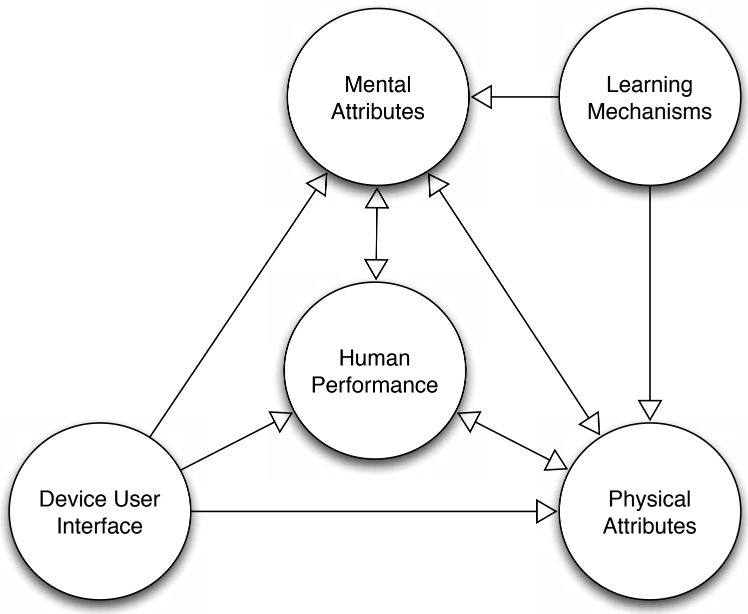

Figure 5-1. Relationship Between Core Themes.
A subset of our seven core themes (*Device User Interface*, *Mental Attributes*, *Physical Attributes*, *Learning Mechanisms*, *Human Performance)* are inter-related and can be used to provide a better understanding of an amputee's experience using our experimental powered prosthesis. E.g., *Device User Interface* (e.g. sensor discomfort) affects *Mental Attributes* (e.g. distraction from discomfort) and also affects *Physical Attributes* (e.g. residual muscle contractions reduce to minimize pressure at sensor site). In this case, *Device Use Interface* does not directly affect *Human Performance*, but affects *Human Performance* via *Mental Attributes* and *Physical Attributes*. An example of *Device User Interface* directly affecting *Human Performance* is signal artifact sent to the controller and altering the mechanics of the ankle.

Core Themes

In the paragraphs below, we highlight the most representative aspects of the user's feedback, separated into the seven themes.

DEVICE USER INTERFACE

Three subjects talked about challenges related to sensor placement or recording an artifact-free signal.

> Subject B: "For me… we found the placement of the sensor was really important for getting good linkage to the [device]."

Subject A: "The only thing that comes to mind would be the sensor location. Try different muscles… and maybe come up with a design of a socket where the sensor was built right into it.

Subject D: "It was also challenging in that there were issues with the sensors and how well they would respond to what I was trying to send in the way of signal. So… as far as quality practice time, that was difficult to come by initially… I walked out of there feeling really crappy when the sensors weren't working right…"

One subject talked about the lack of sensory feedback and suspension issues at the residual limb-socket interface.

Subject D: "… when the heel hits [on your intact side], you have exquisitely good sensors that tell you when that happens… I would want to have more of a definite impact when the heel hits on the [prosthetic] side. That would help me mimic what my left leg is doing better…"

Subject D: "I was keying on how each step feels a little different in terms of how my [residual limb] is inside the socket, how it's moving, and then trying to do fine muscle control when there's a lot of movement in the compartment that it's within… that's one of the variables that makes it more difficult. If I get a lot of movement in there, it's more difficult to sense what the muscle is doing and control it… I don't have a good baseline if the rules are changing as far as contact and feel. So it's a little hard to keep it consistent if the physical feeling is changing somewhat before each step…"

Three subjects talked about the cumbersomeness of the experimental device.

Subject C: "In terms of better testing equipment, I think that a battery powered testing unit would be a lot more desirable and less cumbersome than what you have set up."

Subject E: "Yea… the prosthetic limb was heavier, a little more difficult to move around than my prescribed prosthesis, but it's all minor."

Subject D: "There were encumbrances in terms of weight… and so on, but the basic feel of it was, this is really cool."

MENTAL ATTRIBUTES

Subjects had varying mindsets prior to beginning the study. For one subject, participating in the study was a very new experience and outside her comfort zone.

> Subject B: "When I initially started using the leg, I was a little anxious, a little bit worried… coming into a new lab… there's all this noise and all these parts… just the fact of even taking the foot off my regular prosthesis and putting the limb onto another leg is a little nerve-wracking…. I was actually kind of a little bit scared or timid to try it because I didn't really know, well am I going to fall flat on my face or is this device safe. I was a little bit nervous…"

All subjects experienced some level of mental fatigue during testing, some drastically more than others.

> Subject B: "I [initially] thought of it more as a physical activity, not really one that was going to be testing my mental stamina… I would be so exhausted just because it takes a lot of thought to remind myself, okay, this is what I have to be doing."

> Subject C: "…at first it kind of was mentally fatiguing, but then after I got used to it and flexing that muscle at the right time and releasing at the right time, it was far less fatiguing mentally."

> Subject D: "…concentration is really critical…if I was in a less than stellar mental fatigue scenario, my results were probably not as good."

> Subject E: "I was aware that there was a baseline [mental] fatigue maybe at times… but I never felt as though there was any time in the course of the experiment… that I was either mentally or physically overloaded."

Two subjects thought that their mental fatigue might have affected their ability to activate their residual muscles.

> Subject B: "I don't know if the mental fatigue caused me to not be able to activate my muscle… or not get the timing down."

> Subject C: "… I would get lazy with my leg… my leg gets tired or my mind gets tired, I don't know, and I didn't flex the muscle as quickly as I should sometimes and didn't release at the proper time… maybe more mind control [would have helped]."

Two subjects spent considerable personal time thinking about the experiment and their performance. During testing sessions, they talked about experiences they had outside of the lab and how they could apply it to walking with the experimental powered prosthesis.

> Subject E: "…when I was skiing in Utah and elsewhere, the breakthrough was when I didn't have to look at my feet anymore. Today, because I felt comfortable about using the foot, I actually did that entire session without looking at my feet and just having confidence I knew where I was and knew what it was doing."

> Subject D: "…there was this one time I was walking my dog… I was keying on what my [intact] leg was doing and when in the stride and that allowed me to [figure out] what I was doing a little bit as far as using my muscles [to walk]. It's so instinctive to walk… it isn't something that you delineate, you simply do it, and trying to map it out by consciously thinking about what your muscles are doing isn't the most natural thing to do…"

> Subject E: "…in the later on sessions [with the visual feedback display] where we were trying to get a particular shape to the muscle output, I think that the first or second time I did that I had dreams at night looking into a screen and trying to match it up... so you know, I think there is absolutely no question in my mind that there was a neural process that was going on in the background."

PHYSICAL ATTRIBUTES

Subjects experienced varying levels of residual muscle fatigue. One subject experienced such a high level of fatigue that it impeded his ability to control the device.

> Subject A: "It wouldn't take long before my muscle went, fibrillated to the point where it wasn't controllable… I kind of learned to activate the muscle with the least amount of effort that I could use. Eventually fatigue sets in and then you get a lot of involuntary [muscle activation] and I concentrated on increasing the quite time of the muscle and that worked for a while, but then when the muscle got tired, it just didn't respond like I wanted it to… the first part of the session was the best and then it kind of went downhill."

> Subject A: "In my mind I was thinking if I use this muscle and I practice using it, I could actually improve… because muscle tone is important and if you don't use a muscle it doesn't have the flexibility and control that you would if you were using it on a regular basis."

Two subjects had lower levels of fatigue.

> Subject D: " …[during the walking trials] where I was really evoking and using my residual limb muscle more effectively, there was some [muscle] fatigue involved, but it only took I think just concentrating a little bit harder in order to make those muscles do what I'd learned they can do."

> Subject B: "During the first few [sessions], I remember [my muscle] being pretty sore where I'd have to take the leg off, massage the muscle, and it was very physically tiring… I felt like I was pumping iron with my leg, it was just that tired."

Some subjects demonstrated involuntary residual muscle activations when they walked. It was not clear if they were reflex mediated or a result of a chronically learned muscle patterned. In one subject, he felt that stretch reflexes caused the involuntary muscle contractions. This subject often talked about this phenomenon.

> Subject E: "I know there are certain postures I can put my residual limb into where…depending on the load on the socket… the stretch receptors in my leg are firing as a result. I get a [residual muscle contraction] when I don't really want to…"

> Subject E: "It is definitely the stretch receptors. There isn't any question in my mind that it's an autonomic response to the unbalanced condition or the

out of control condition that you get either early elicitation or drastic late elicitations… caused by stretch [receptors]"

Subjects had varying degrees to which they could feel their residual muscle contracting inside their prosthetic socket during walking. On one extreme, subjects could barely sense their muscle contractions. On the other extreme, one subject was able to sense how his "foot" was moving based on his residual muscles contracting.

Subject A: "…I couldn't feel if the muscle contracted intentionally…could have been involuntary, probably was."

Subject B: "I tried to play around with activation timing, but it was difficult to know when to do it. When I'm seated, I can feel [my muscle] contraction, but when I'm walking, I don't feel it contract."

Subject D: "…I can't imagine not being able to at least have some sense of the [residual muscles]… [Walking with your device], I can sense what is going on between the ground and the prosthetic. I think it has to do with why people get phantom pain. The nerve is still intact, and you have a sense of where your big toe was and your heel was… you can feel the shape of your foot and you do get a fair amount of input from how your socket impacts against your [residual limb]… I can move each part of my foot with the residual muscles that I have. I am envisioning or feeling a foot."

Three subjects felt that it was challenging to sense and/or control the midrange residual muscle activation during walking.

Subject B: "I can't tell graded contractions, only whether it is 100% vs. 0%. That's why I contract all the way sometimes, to make sure I'm still getting power. I'm not sure whether the muscle was engaged unless its 100% engaged."

Subject E: "It's the zone between 25 and 75% [of maximum contraction] that center zone there where the whole sense of… is [the device] doing what I want it to do? ...There are some [muscle activation] patterns that are

> relatively easy to fall into that are only really eliciting an eighth or quarter of what I'm capable of from a muscle aspect."
>
> Subject D: "I think it was a really good thing for me to try to involve the muscles throughout the gait rather than just trying to fire [the muscles] at a certain point and that is something that physiologically is a little difficult."

Subjects had varying sense of body motion during walking. One subject had difficulty maintaining awareness of body motions and gait symmetry while another subject clearly sensed differences between intact and prosthetic limb motions.

> Subject B: "I don't know what feels right, so it's hard to know from step to step what it's supposed to be. I know there are mistimed steps, but… I have no sense what is right. Well, I know it's not right when I get jolted forward or when it's not enough power… because [the prosthesis feels] heavier. It's hard to say. I feel like I don't even notice the difference or similarity between right and left sides."
>
> Subject D: "I'm trying to mimic what I'm doing with my [intact] leg's muscles with my residual limb muscles. It's a fairly smooth, subtle motion... intimate control… it's easy enough to get a balanced feeling [between my intact and prosthetic] side."

LEARNING MECHANISMS

Subjects wanted more direct instruction on how we wanted them to walk with the powered prosthesis. We had instructed them to walk on the treadmill using the powered prosthesis as they felt comfortable, but they wanted specific guidance on how exactly they should behave during walking.

> Subject A: "…it takes a few sessions to understand what kind of outcome we're looking for… once I got that hang of that part I could start learning different ways to achieve that outcome…"

> Subject E: "…I think from a learning perspective I went through a fair amount of learning. A few of the areas at least initially, weren't entirely clear to me in terms of what we were trying to accomplish."
>
> Subject D: "The visual aid was really good, but I think I also… really needed to know better what you were after."

Most subjects thought that the visual feedback display during walking was helpful and provided them with a better understanding of how they were using their residual muscles to control the device.

> Subject A: "I had a desire to learn how to use [the powered prosthesis] and I definitely felt that once we got the visual feedback monitor, that kind of gave me a lot better understanding of it… I got a little more enthusiastic because… I could see what I was doing and I could change what I saw on the screen by how I dealt with the leg."
>
> Subject D: "…I think the visual feedback, the last session where I was able to make [the control signal] stay in the [target] and understand what I was doing to make it stay in that [target] was really quite satisfying… I was able to control the thing like it was part of me because I was able to do things and see the effect and feel the effect walking as well. Seeing it visually is very nice to have validation, but also feeling it is a great type of validation as well."
>
> Subject C: "After you put the computer screen up, where I could actually see what was going on [with my muscle activation], it was a lot better as far as getting fatigued because I could see how much I had to flex and when I had to release and that was good as far as not getting tired so much, because then I knew what was expected."

Although the visual feedback helped with subjects' understanding of the controller, one subject in particular felt that she could have benefitted more from a better understanding of how the entire system worked. This is contrary to another subject who was well versed in human biomechanics and immediately understood how to control the device using his residual muscles.

> Subject B: "I think I respond really well to visual stuff and positive reinforcement… also being very repetitive with instructions [on how I can use my residual muscles to walk with the device]… I think that was very helpful for me, I could never have enough… just breaking it down… I never really think about when I'm walking and I place my heel, this muscle is activated and that's when I need to activate that muscle… it just doesn't click for me very much, so being able to repeat those instructions over and over again helped kind of link it up physically."

> Subject B: "…seeing every part of the device kind of breaking it down of how they all interact, or just maybe starting from square one, just walking with your regular leg and saying, okay, this is incrementally what is going to be changing using the new powered prosthetic, could be helpful."

All subjects thought they would have continued to improve their ability to walk with the powered prosthesis if they had more exposure time.

> Subject D: "One thing that does keep coming back is this does take practice. Because I think I'm getting better at modulating how I use the residual muscle so that I can do it incrementally rather than just blast on and blast off, but it does take some practice to do that… if I was able to have more time [practicing], it would really come through as far as showing me… what sort of an advantage it would be."

HUMAN PERFORMANCE

There were two subjects who clearly didn't learn to walk as well with the powered prosthesis compared to the other subjects. When they talked about their performance using the device, they focused on their ability to navigate the visual feedback display rather than their ability to walk with the device.

> Subject A: "I think I was pretty successful and our last few sessions, I think I was getting a lot more confident and I think I was staying in the [visual feedback target zone] a lot better and I knew the reasons why I was going out of the [target zone] so I was able to control that better until my muscles got tired."

Subject A: "I had a premature muscle activation and it would throw me out of the [visual feedback target zone], but if I relaxed and fired [my muscle] at the right time… I could stay in the zone."

Subject B: "I felt like I did see some progression as far as me being able to learn how to use [the powered prosthesis] over the course of time… the first two sessions I was pretty much like, oh my gosh, I'm never going to learn this, it's so difficult… by the end… when we had the visual feedback stuff I felt like okay, I can do this… I'm making progress. I can see what my actions are linking up to what's on the screen and I'm kind of learning the process a little bit more."

There were two subjects who were clearly more advanced in their ability to walk with the powered prosthesis compared to the other subjects. When they talked about their performance using the device, they focused on their ability to control the behavior of the prosthesis.

Subject D: "The further we went through this process the more quality practice time I had with it and also I was able to train my residual muscles to do things with a little nuance and fine control… trying to do something that is more subtle like an easier push or a timed push does take practice, but the process was quite exciting because I think at the last few sessions… I could get good response from the limb and I was then able to really do quality practice time on it… just the concept of having my body more balanced in terms of propelling myself walking was a very good feeling... that is empowering definitely. It's nice to not have the rest of your body make up for what your calf muscles aren't doing…"

Subject D: "…it's a little bittersweet… I felt our very last session was one where I was really controlling my muscles in ways that I could customize how I was walking. I was getting to the point where I could really, I think, control the thing in a way that was accurate… at the edge of really using it in a quality way… I was becoming a lot more effective and really learning how to use it."

Subject E: "[My] sense of control is getting much improved… I feel like I have better control over it… It's doing what I want it to do when I want it to do it and I'm not feeling as though I'm having to consciously control it, but sort of more ad-hoc."

Subject E: "I mean I've gotten to the point now where I actually think I could walk around without a safety harness… I've gotten better at [recovering from perturbations]. There was a time that I didn't feel as though I had a whole lot of control over that. I mean I thought I had control, but the reality was that it was hard after a misstep, it was hard to recover…"

The two most advanced subjects recognized how the experimental powered prosthesis behaved in a way that was advantageous compared to their prescribed (passive) prosthesis, but they also emphasized the importance of reliability.

Subject E: "[Your device] really helps in the process of walking. I noticed an immediate difference in terms of the stability of the transition zone [where I'm transferring the limb from the rear part of the step to the fore part of the step]…. but fundamentally I hadn't even realized until I had the solidity of [your device] how much that makes a difference in the transition. There is sort of a wobble…sort of a roll over zone in the [prescribed] prosthetic foot that is different from [your device]. When I use [your device], I get more of a sense of control."

Subject E: "They're not even the same. I don't get any help from my prescribed prosthesis really what-so-ever. There is a little spring action, but it's mostly a shock absorber. That is not anything that really gives me anything at the end of my step… where [your device] really does… so it's cool."

Subject D: "… the overall feeling is that [when I'm walking] with a prescribed prosthesis, it's a dead prosthetic, you're rolling over it. If you have this hooked up, there is a definite parallel to having regular muscles… This powered prosthesis is very much like [a flex foot], but on steroids. It responds to my muscle input very, very quickly."

Subject E: "…my [prescribed prosthesis] is reliable and I can count on it…there is a very practical aspect to a prosthetic limb and it has to be very very very dependable."

Subject D: "… the reliability is quite important to me. That is a very high end priority."

SIGNIFICANCE

Subjects felt that this research is important for advancing prosthetic technology.

> Subject B: "… having gone through this I think it's really impactful and it's something that is needed for an amputee to see that there's progress being made on prosthetics…"
>
> Subject D: "… I think that the intent of all prosthetists and the people who design the equipment for amputees is to come up with the best version of whatever they are replacing… and this [powered prosthesis] really adds a lot to what a prosthetic can be… if the framework is established within what we did, that one can control a [powered prosthesis] with these signals that are derived from the nerve tissue and the residual muscle, the mechanical ability to do this with technology… is going to come…"
>
> Subject E: "… from a scientific enabling standpoint very very interesting, very very worthwhile… in terms of what we could be capable of in the realm of technology, having an actively controlled prosthetic ankle would make a huge difference [for the amputee community] and… from the standpoint of neuromuscular control in prosthetics, in some ways [it's] an area that's been underutilized, underfunded perhaps and underexplored."

Subjects felt that having a powered prosthesis that was controlled similarly to the experimental powered prosthesis would enable them to do more activities and perform the activities they can already do, but better.

> Subject D: "I could see [something like this powered prosthesis] improving quality of life. I think the ability to do things physically, as far as different types of activities, would make it so being a below knee amputee would be pretty much a non issue… it would not be a handicap per se…I think I'm reasonably involved in athletic things, [so] for me that would allow the total pursuit of that kind of thing whereas there are limitations to what I can do now... having that control of your own body is definitely empowering and therefore gives you better quality of life."
>
> Subject E: "I know there are some activities I struggle with, dancing for instance is very difficult…so yea, I think in general, articulated prosthetic limbs… are enabling. They give amputees an opportunity to have a much broader range of activities they can get involved in and… that increases their self-confidence and their ability to function in the world."

EMBODIMENT

Three of the five subjects talked about how using the experimental device was empowering and made them feel more “complete”.

> Subject B: “It was like a sense of rebirth… I don’t know if that’s being dramatic, but it was just so cool to be able to stand up on both of my ankles, and being able to visualize and see the foot moving up and down and knowing that was being done as a result of my limb…my muscle contraction… being able to have a limb that responds to your own muscle contractions and is able to give you power that aligns with how you want it to work, versus just something that’s hanging on your leg doing what it wants to, its something that’s really liberating for someone whose had part of them taken away. It gives them a sense of ownership and pride.”

> Subject E: “Initially when I got involved in the study, the initial experience was that it was exhilarating to have the ability to control the ankle… I felt like a science fiction thing, I was The Terminator… I was Luke Skywalker with a new hand, I felt like I was Avatar with a new body. I mean there was a whole lot of what I would describe as scientific speculative thoughts that went through my head as a result of one, seeing [your powered prosthesis] and then two, getting to the point where I could control it… When I got done with the experimental session, taking off the foot and replacing it with my prescribed prosthesis made me very aware that there was a decrement in terms of the quality of how I felt… I got attached to [your device].”

> Subject D: “It was a very enabling type of thing to have it hooked up and when I wanted to make it do something, it did it. I have been an amputee since 1970… that’s the majority of my life. And having it hooked up so it worked was exciting… it basically felt like I had more freedom and frankly, it felt like I was younger, but also more complete… I really felt that I was more in control of a complete body… It’s not a perfect mimicking of the other side… it just has a lot of the elements of a limb that you can really control. You don’t have everything you used to, but it’s a really neat recovery of a lot of the things you used to have…”

Narratives

SUBJECT A appeared to walk with a little more caution compared to the other subjects when he walked into the lab. It didn't look like he was unaccustomed walking with his prosthesis, but he took shorter strides and stepped up and down curbs more deliberately. One day when I was walking outside of the lab with him, I noticed that he asked to stop for a break about every ten minutes or so. His prosthetist suspected there might have been a circulation issues with his residual limb that prevented him from walking comfortably for an extended time.

Of all the subjects, he experienced the most residual muscle fatigue when he walked with the experimental powered prosthesis. Although he felt that he was able to learn to walk with the powered prosthesis, he was very aware that his muscle fatigue severely limited his learning capabilities. Compared to other subjects who might have had ten walking trials in a typical testing session, he averaged two walking trials per session before his residual muscles began to fatigue. Even with long breaks between walking sessions, he wasn't able to recover from his muscle fatigue. It took us by surprise on his final testing day when he asked if he could opt out of performing any non-walking trials where he would have to use his residual muscles. He wanted to save all the strength he had in his residual muscles for walking.

Subject A walked with odd gait patterns with the powered prosthesis. It seemed like his general strategy was to alter his gait pattern as a means to change his residual muscle activation pattern or to minimize involuntary or undesirable residual limb muscle activation. One time, he adopted a marching-style gait where he would land with a flat foot instead of with a heel strike in an attempt to minimize his residual muscle activity following heel strike. Another time, he adopted a gait where he flexed his knee considerably during early to mid stance, then fully extended his knee during late stance in a spring-like action to elicit a certain residual muscle activation pattern for push off. He was able to walk with a less awkward gait pattern with verbal corrections, but his natural tendency was to revert to odd gait patterns when walking with the powered prosthesis.

Subject A's perception of how well walked with the powered prosthesis was much different that what we expected. During the visual feedback portion of the study, he would say that he was improving or that he walked well during a certain trial. But if we had to guess, we would have thought that he was having an awfully difficult time controlling the prosthesis. Adding visual feedback seemed like it was a game changer for him. He told us many times that the visual feedback made him more enthusiastic about walking with the powered prosthesis and we did notice this change. During the no visual feedback portion of the study, we got the feeling that he was indifferent to returning for the next testing session. However, during the visual feedback portion of the study, he seemed eager to come back.

SUBJECT B lost her leg due to cancer when she was just a child. She grew up with her prosthesis and doesn't remember what it was like to walk with an intact leg. When she first came into the lab, she was so amazed by the lab and the research that all the equipment that we had set up. She had never been involved in experiments similar to this before. She was so curious about what the study entailed and really appreciated learning little bits and pieces about our prosthetics research and the science behind everything. She was curious, yet reserved.

During her initial testing sessions, she was a bit concerned about what affect using her residual muscle to walk with the device would have on her residual muscle. She wondered if it was okay for her to be using her muscle like she was in the study, if it was healthy. She was cautious about her residual limb because she had difficulty in the past with socket comfort. With her prescribed prosthesis, she wears a Pe-lite liner over a silicone liner (which is atypical) because without the Pe-lite liner it is too painful for her to walk. Her prosthetist couldn't quite figure out why wearing a silicone liner alone was painful for her and it remains a mystery.

When we asked Subject B how it felt to walk with the powered prosthesis, she had trouble comprehending what the powered prosthesis "should feel like". When we asked her questions about how the powered prosthesis felt compared to her prescribed prosthesis, she had difficulty comparing the two. However, one day when we were walking down the stairs to the lab, she commented that after her last testing session, when she was walking or running with her prescribed prosthesis, that it felt "dull". So it seems that she did have at least some sense that the experimental prosthesis gave her some advantage during walking. Even though Subject B was perhaps the least confident walking with the powered prosthesis, she told us many times that participating in this experiment was life-changing situation and she always expressed gratitude in having the opportunity to be involved in the study.

SUBJECT C had his amputation due to intractable ankle pain. He did a lot of hiking and walking. Even though he was used to walking for long distances, his natural tendency walking with his prescribed prosthesis was to walk with his torso tilted a little forward. During walking trials with the powered prosthesis he always held onto the handlebars of the body weight support system. When we asked him about his prescribed prosthesis and how other people perceived him, he was quick to say that nobody can usually tell that he is an amputee when he walks because he doesn't walk with a limp. There were several times throughout the study where he told us that everyone is surprised when they find out he is an amputee because he walks normally.

When we asked him about what his initial impressions were with the device, he used the word "strange". He said it was a "strange feeling" and something that he wasn't used to. Although it was hard to interpret what he meant by "strange", he did indicate that he felt the powered prosthesis helped him walk in a positive way because using his residual muscle to walk did "help the foot move correctly and with less fatigue". Something odd about Subject C was that during his last interview, he talked about using his residual muscles to control his prescribed prosthesis. It almost seemed as though he was confused about how the experimental powered prosthesis and the myoelectric controller actually worked. He said that after he starting participating in the study, he started to contract his residual muscle when he walked with his prescribed prosthesis because he realized that contracting his residual muscle could "help in the process of walking". He said that by contracting his residual muscles during walking, there was less pressure on his residual limb and it felt "better" to walk in that way.

Overall, it was a little difficult to figure out how Subject C felt about the powered prosthesis. He did tell us that he enjoyed and appreciated being included in the study. In fact, he didn't mind driving five hours to get to and from the lab on testing days. And although he wasn't always clear in communicating how he felt, he ended his last interview with a response we didn't need to interpret. He said, "I know that eventually you will be involved in a new foot that will be able to be marketed to amputees that is better than what is out there now." So it seems that he knew that this experimental prosthesis had something new to offer in prosthetics.

SUBJECT D had his amputation when he was a teenager as a result of a very severe electrical burn. He talked a lot about how his amputation changed his life trajectory and how crude the prosthetics were in the 1970s. He had a very active lifestyle and did a lot of mountain biking. Even though he had an extraordinarily stressful and busy life, he always made time to come into the lab.

Subject D was unique from other subjects in that he paid especially close attention to how his body felt when he was walking with the powered prosthesis. His was able to tune into how he used his intact muscles when he walked as a strategy for teaching himself to walk with the powered prosthesis. After testing trials where he really felt like he had good control, he talked about how much more symmetric the powered prosthesis made him feel compared to his prescribed prosthesis, how balanced his left and right sides felt. He also talked about having the sensation of moving a small foot inside his prosthetic socket and having the sensation of pushing off with an actual foot when he walked with the powered prosthesis. Interestingly, he was the only subject who talked about experiencing phantom pain occasionally since his amputation.

Subject D was more emotionally invested in the study compared to other subjects. In general, it seemed like he spent a lot of time thinking about the study on days when he didn't come in for testing or later in the evening on testing days. His perceived success during a testing session would often determine his outlook for that day. This meant that on days where we had issues with sensors and it was difficult to get a record a reliable control signal, he felt "really crappy". He would say that he wasn't sure the sensors weren't working right or if "he wasn't working right". The flip side was that when we didn't have sensor issues and he was able to control the prosthesis confidently and performed really well, it was "very satisfying". During his last interview he said that it was "depressing" to think about the study ending because the experience had such a huge impact on his life. If I didn't know better, I would say Subject D had a vested interest.

SUBJECT E lost his leg as a result of a single-engine plane crash. He had a really confident and adventurous personality. He also seemed like he had fun playing a troublemaker. He came into the lab with a background in human biomechanics research and in fact, he completed his doctoral degree on research related to human work performance and electromyography. He had so much insight into the experiment that during some of his testing sessions he would talk about the controller and what he thought could be improved.

He was comfortable walking with the powered prosthesis from day one. After the very first walking trial, he asked if we wanted feedback then he threw his hands up as he exclaimed “Awesome!”. He continued on to say that walk with the powered prosthesis was the first time that he actually felt like he wanted to run. When he sat down after his first walking trial, he couldn’t stop talking about how much better walking with our experimental prosthesis felt compared to his prescribed prosthesis. He compared the powered prosthesis to a racecar that you could tune and crank the power “up, up, and up!”

Between walking trials when I asked him what things he would like to try with the powered prosthesis, he would go through a long list… playing a drum set and being able to use a kick drum, spiking a volleyball so he could be more competitive at his game, squatting down to remove a heavy food tray from the oven without worrying about burning his arms, walking up and down the stadium stairs without handrails without worrying about his balance, walking in a “squishy” (foam) pit. During testing sessions he would often recall something he did over the weekend where having the experimental prosthesis would have been beneficial.

Subject E acted like walking with the powered prosthesis was second nature. During one of the testing sessions, he talked about letting the process of walking be “autonomic”. He said “I was deliberately distracting myself with music, I was humming yellow brick road.” He was the only subject who didn’t tell us that the visual feedback helped him understand how to control the powered prosthesis. In his interview he said “…I suppose if you are threading a needle a thousand times after a hundred it probably gets to be a little bit of a bore…” Because Subject E felt so bored walking on the treadmill at the same speed day in and day out, he tried to get away with doing as much as the hardness and treadmill would allow. He told us about the “games” he played when he was walking. One of them was simply counting in his head how many steps he could take before he had a misstep, where the prosthesis didn’t do what he wanted it to do. Or, he would purposely throw himself off balance using the controller, then see how many steps it took him to recover from the perturbation. Subject E appeared to have learned to walk with the powered prosthesis much more quickly than the other subjects, and it really did feel as though during testing we were holding him back. On his very last trial we let him walk at faster speeds and each time we increased the treadmill speed he would say “faster, faster, faster!”.

Word Frequencies

The word frequency visualizations that we generated for each subject (Figure 5-2) are very interesting because the results align with our direct observations of how confident the subjects were when they walked with the powered prosthesis. Based on direct observations of each subject during the first testing block (no visual feedback), the rank order from lowest to highest confidence when controlling the prosthesis was: 1) Subject B, Subject A, Subject C, Subject D, and Subject E. Accordingly, the two subjects who were the best at controlling the experimental powered prosthesis (Subjects D and E) used the word "control" more than the other subjects and did not use the word "think" as frequently as the other three subjects. The two subjects who had the most difficulty walking with the prosthesis (Subjects A and B) used the words "contract" or "contraction" more frequently. The subject who had the most difficulty with residual muscle fatigue (Subject A) used the word "fatigue" more than the other subjects. Subject E was noticeably the most confident walking with the experimental powered prosthesis and the two words that dominated his interview responses were "control" and "foot".

Figure 5-2. Word Frequency Visualizations.
Font size and darkness scales with word frequency. Minimum frequency=5, Maximum words=30. Rank of subjects by increasing confidence walking with powered prosthesis is B, A, C, D, E. More confident subjects used the word “control” at a higher frequency. Less confident subjects used the word “contract” or “contraction” at a higher frequency. The frequency of the word “think” generally decreases with increasing confidence.

DISCUSSION

The results that emerged from the subjects' interview responses and verbal feedback (via the seven core themes) were very insightful because they helped to identify specific factors that might significantly affect the user's experience with a powered lower limb prosthesis under continuous myoelectric control. Below we discuss some of these factors, which include: signal artifacts, sensor placement, residual muscle fatigue, involuntary muscle activations, feedback mechanisms, and embodiment.

Signal Artifacts

During our initial testing sessions, we had a relatively high occurrence of signal error due to poor sensor reliability. Sometimes sensor reliability was poor because we were not familiar with recognizing signal characteristic indicative of the beginning of sensor failure and we could resolve the issue by simply replacing the sensor. More often however, sensor reliability was poor due to artifacts generated from micro-motions between the skin and the electrodes. The presence of signal artifact created from micro-motions had a strong negative affect on Subject D. These artifacts were especially hard to minimize via filtering because their frequency range overlapped the frequency range of physiologic signals. We had a more difficult time recording artifact-free signals with our experimental powered prosthesis compared to a prescribed prosthesis because the dynamics at the at the limb-socket interface of the experimental prosthesis were more volatile due to the nature of our controller. As the study progressed, we were able to modify the limb-sensor-socket interface to minimize signal artifacts generated from micro-motions, which increased sensor reliability. Factors that we believe affected the magnitude and/or frequency of signal artifacts were residual limb contour (smooth skin vs. skin with creases), liner material properties (alpha liner vs. silicone liner), liner condition (old and “stretchy” vs. new), and suspension (e.g. “tight” vs. “loose”).

Sensor Placement

In addition to recording a clean residual muscle signal, another challenge was being precise when choosing the sensor placement over the muscle. For the two subjects (Subjects A and B) who had noticeably lower residual muscle volume and/or muscle tone, we found that being precise with sensor placement was critical. Their residual muscles required more attention to palpate because their muscle did not protrude above the skin. We also noticed that shifting the sensor placement slightly (i.e. 5-10 mm), while still over the “belly” of the muscle, yielded an altered characteristic (i.e. base) residual muscle activation pattern during walking with the powered prosthesis. The importance of sensor placement with Subjects A and B is that certain base activation patterns were easier for them to adapt than others. Sensor placement with subjects who had larger residual muscle volume and/or muscle tone (Subjects C, D, and E) was not nearly as sensitive.

Residual Muscle Fatigue

Aside from overall sensor reliability and sensor placement, another factor that affected the quality of the myoelectric control signal was muscle fatigue. We had two subjects (Subjects A and B) who developed residual muscle fatigue relatively quickly, which limited the duration of each trial. The duration of the walking trials seemed to be important because several subjects mentioned that they took a couple minutes to get into the rhythm of walking. In fact, two subjects (Subjects D and E) said that they felt like we cut their walking trials short and thought they would have benefitted from longer walking trials. Subject A experienced the most muscle fatigue. Oftentimes he would only be able to walk for one or two short (3 minute) trials before fatigue set in and his control signal became unmanageable. Persistent muscle fatigue might have caused amputees to adopt a control strategy where minimizing muscle fatigue was the primary goal rather than achieving greater push off or walking with increased stability. Not surprisingly, we observed that the two subjects with the weakest residual limb muscles (Subjects A and B) struggled the most to control the powered prosthesis. We believe that these subjects would have benefitted from conditioning their residual muscle in order to build muscle volume and/or muscle tone prior to walking with the experimental powered prosthesis. Subject A felt really frustrated because of how quickly his muscle fatigued during testing and told us during his last interview that he thought he would have had a much easier time learning to walk with the device if he would have been able to train his muscle beforehand.

Involuntary Muscle Activation

In at least one of our subjects (Subject E), dynamic changes in limb-socket loading appeared to generate a reflex response. Involuntary muscle activations such as muscle stretch reflexes are problematic because they can result in a large residual muscle activation signals and thus, large perturbations during walking with the powered prosthesis. We suspect that certain physical interactions at the limb-socket interface may have caused muscle stretch reflexes such as concentrated pressures at the sensor site. Subject E was able to figure out how to avoid involuntary residual muscle activations by avoiding certain postures, however that was not ideal. Ideally, the user should be able to walk with any gait pattern that they choose without eliciting a supposed muscle stretch reflex.

Feedback Mechanisms

One reason why we think that verbal instruction and correction were important factors is because the subjects did not have a good sense of their prosthetic ankle position and foot placement during walking. For many subjects, we had to play back videos of their walking trial or mimic their gait pattern in order to convince them of their gait pattern. For example, during one walking trial, Subject D was completely unaware that he was landing on the ball of his foot instead of heel striking. In another example, Subject C was completely unaware that he was flexing his knee so much during mid stance. When we told him that his prosthetic-side hip was dropping down during stance, he was able to pay attention to his hip in subsequent trials and walk with a more symmetric gait. Later he told us that walking with his hip was level actually felt better. If subjects would have been able to see their gait pattern and posture in real time, we might not have needed to give them so much verbal guidance.

A few subjects told us that one of their strategies for learning how to walk with the powered prosthesis was to look down at their feet in an attempt to get more information about the prosthesis behavior. This suggested that the sensory feedback available to the subjects at their residual limb-socket interface and the kinematics of their proximal joints were insufficient for understanding how the prosthetic foot was behaving. Although subjects felt that looking at their feet provided them with a better sense of timing, they said that they felt more comfortable when they had their head oriented forward instead of tilted down. We did notice that subjects who looked down at their feet while walking were not able to maintain their head position for more than a few strides before tripping. Even the most confident subject (Subject E) said that it would have helped to see what his feet looked like when he walked, especially in the early learning phases.

Subjective responses about the visual feedback display suggested that subjects thought that seeing their residual muscle activation patterns in real time was a useful learning tool. Some subjects felt that the visual feedback helped them understand how the prosthetic-controller system worked. Additionally, some subjects said that the visual feedback helped reassure them that the experimental powered prosthesis, mainly the sensor, was working properly. Another important aspect of the visual feedback was that it engaged the subjects. Many subjects told us that the visual feedback made it more exciting to walk with the powered prosthesis. They said it was "neat" and "fun" and made them "enthusiastic". These results demonstrated that the visual feedback was a useful learning tool in multiple ways. The visual feedback 1) helped teach the subjects how to use myoelectric control, 2) provided confidence that the experimental device, particularly the senor, was working properly, and 3) motivated the subjects to continue learning.

Embodiment

Perhaps the most compelling result of this study was that the subjects experienced a level of prosthetic embodiment when they used our powered prosthesis that was far beyond their prescribed prosthesis. Interestingly, the level of embodiment that the subjects felt when using the powered prosthesis was not necessarily related how well they were able to walk with the device. For example, even though Subject A had a relatively difficult time understanding how to walk with the device, during her interview she told us that when she used her residual limb muscles to move the prosthetic ankle for the first time, it gave her a sense of “rebirth”. The level of embodiment and empowerment that Subject E felt when he walked with the powered prosthesis made him feel like he had a “new body”. Subject D said that being able to control the prosthesis made him feel more “free”, “younger”, and more “complete”. Another subject said that when he walked with the experimental powered prosthesis, it “feels and looks more natural than the iWalk”. These subjective responses suggest that continuous myoelectric control of one’s prosthesis might have a considerable affect on how amputees rebuild their “sense of self” following amputation.

CONCLUSION

The level of prosthetic embodiment that a volitionally controlled powered prosthesis can provide an amputee is far beyond that which a passive prosthesis or a state-based powered prosthesis can provide. It is conceivable that in the near future with advances in sensor technologies and socket suspension technologies that we will be able to design commercially viable powered lower limb prostheses that amputees can control directly using their residual muscles.

Chapter 6: Discussion and Conclusion

The main purpose of this dissertation was to conduct a feasibility study on using residual limb muscles for continuous proportional myoelectric control of a powered lower limb prosthesis during walking. The goal was not to create a commercially viable device or controller, but to demonstrate a proof of concept. A great deal of future work will need to be done to transition these results to clinical applications. The development of future technologies, such as prosthetic socket and liner materials that reduce stresses on the residual limb [96], intramuscular myoelectric sensors [29, 30, 87], advanced batteries, and miniaturized electronics, will certainly have a profound impact on robotic lower limb prostheses that reach the commercial marketplace. There is still a need, however, for basic science research on how lower limb amputees adapt and use powered prostheses before the field can determine which types of controllers and prostheses will have the most success in the long term.

My first study (Chapter 2) was a critical first step because it suggested that it was feasible to utilize residual muscles of transtibial amputees for continuous myoelectric control of a powered prosthesis during walking. The primary result of the study is that I demonstrated that it was possible to record robust residual muscle activation signals using surface electrodes at the limb-socket interface during walking. Not only did I find that it was possible to obtain sufficiently high-quality residual EMG during walking, but I also found that residual muscle activation patterns were consistent across strides. Another finding from my first study was that the activation patterns of residual muscles were significantly different from the activation patterns of corresponding intact muscles during walking. Residual EMG patterns were different than intact EMG patterns, but also highly variable across amputee subjects. These findings indicated that post-amputation, the amputees learned to adapt their residual muscle activation patterns (i.e. motor plasticity) in a manner specific to walking with their own prescribed prosthesis. This was a positive result because it implied that amputees would be able to further adapt their residual muscles to control a powered prosthesis if their residual EMG was linked to prosthetic ankle function.

My first study was significant because it was the first time that the signal quality and signal variability of residual muscles of transtibial amputees (or transfemoral amputees) were quantified systematically during walking. Additionally, it was the first study to explicitly present EMG data from lower limb residual muscles during walking. Previous studies have used residual EMG for state-based myoelectric control of lower limb prostheses, but failed to show important characteristics of the residual muscle control signals such as signal quality and signal variability [19-21, 23]. Understanding these signal attributes (i.e. quality, variability) was a critical first step in pursuing the use of residual muscles for myoelectric control not in a state-based, but in a continuous manner during walking.

The goal of my second study (Chapter 3) was to design an experimental transtibial powered prosthesis that I could use as a platform for testing a proportional myoelectric controller via residual muscles during walking. I designed a powered prosthesis with a freely articulating ankle in the sagittal plane, actuated using pneumatic artificial plantar flexor and dorsiflexor muscles. I used the same methods from my first study to interface the EMG sensor with the residual limb muscles to record activation signals at the limb-socket interface. One feature of my experimental prosthesis was that it was compatible with the amputee user's prescribed prosthetic socket. This meant that I was able to reduce overhead time and cost associated with designing a custom prosthetic socket for each individual subject. Another feature of my experimental prosthesis was that the ankle torque output range and set-point ankle stiffness were easily tuned by reconfiguring the pneumatic artificial muscles. This was an important feature because it allowed us to be selective about the passive behavior of the device. Although I designed the experimental device for the purpose of implementing a continuous myoelectric controller during walking, the reconfigurable nature of the device also makes it an ideal test bed for examining the affects of ankle stiffness on the performance of locomotor and discrete tasks, which is a high interest area in designing transtibial prostheses that remains somewhat of a mystery.

To conclude my second study, I pilot tested the my experimental powered prosthesis and demonstrated that it was possible for a transtibial amputee to adapt his residual muscle activation patterns to dictate ankle mechanics during walking via continuous proportional myoelectric control of plantar flexor muscle force (i.e. plantar flexor torque). Within a very short time (i.e. 30 minutes treadmill walking), the amputee adapted his residual muscle activation in order to generate symmetric ankle power between his intact and prosthetic sides. This result was extraordinarily exciting because it was the first study to successfully implement continuous proportional myoelectric control of a powered lower limb prosthesis during walking in the absence of a finite state machine control framework. I recognize that the use of state-based controllers is necessary to ensure the safety and reliability of powered lower limb prostheses that are being developed for the commercial market. However, I have no intention of developing my experimental powered prosthesis for clinical applications. I was simply trying to demonstrate a proof of concept.

In my third study (Chapter 4) I expanded on my pilot testing from my second study and demonstrated that five amputee subjects were able to adapt their residual muscle activation patterns to walk with increased prosthetic ankle power (over their prescribed prosthesis), but only when they were presented visual feedback of their control signal in real time. Without visual feedback, subjects did not naturally settle on residual muscle activation patterns that resulted in increased prosthetic ankle power. This result may be related to the limited amount of practice the subjects had (i.e. 3-4 hours total walking time) or it may be a fundamental advantage in learning. The motor learning literature has repeatedly demonstrated that expert feedback can aid in learning novel motor tasks [81-83, 97-99]. Visual feedback about key biomechanical parameters during walking might allow the amputees to experience a greater state space of motor parameters and learn at a faster rate.

Another explanation for why my subjects did not walk with increased ankle power using the experimental powered prosthesis might be that they simply chose not, even if they were capable. In fact, that's the precise purpose of the controller- to allow the subjects to use their prosthetic ankle as they want. It is very probable that in the absence of visual feedback during walking, the amputees were "optimizing" walking parameters such as stability (maximizing) or residual muscle work (minimizing), both of which could conflict with the objective of matching intact side ankle power or ankle work. If this was the case, then providing subjects with visual feedback of their control signal and instructing them to match a target activation region was telling them they should "optimize" ankle power. The idea that amputees would place more or less importance on certain walking parameters (e.g. stability, power) makes sense in the context of walking speed. For example, if I asked my subjects to walk on the treadmill without visual feedback at 1.0 m/s, then after 2 minutes gradually increased the belt speed to 1.6 m/s and asked them walk for another 2 minutes, they might naturally reweight their gait parameters and maximize ankle power, even at the cost of stability (which they weighted more heavily than ankle power at 1.0 m/s). Now, at 1.6 m/s they might be using the experimental prosthesis to generate more ankle power than their intact side when at 1.0 m/s they were content with generating less ankle power than their intact side.

In my fourth and final study (Chapter 5) I used subjective response data to assess the amputee users' experience walking with my experimental powered prosthesis using continuous proportional myoelectric control. This study was important for two reasons: 1) It revealed the strengths and weaknesses of my experimental myoelectric powered prosthesis, and 2) It highlighted the potential physical and emotional advantages that continuous myoelectric control might have over purely state-based or passive devices. One of the most compelling results of this study was that despite its challenges, continuous myoelectric control provided my subjects with a sense of empowerment and embodiment that is likely unique to neural control via residual limb muscles.

One general limitation of my study is that I only recruited five amputee subjects to walk with the experimental powered prosthesis. As a result, I did not have evenly distributed amputee characteristics (e.g. age, gender, reason for amputation, time since amputation). However, my subjects were fairly diverse in their characteristics and certainly their behaviors were vastly different. So although I had a small number of subjects, I felt that they provided a fair representation of the larger amputee population (excluding diabetic amputees).

A second general limitation of my study is that I did not design custom prosthetic sockets or liners for the amputee subjects to accommodate the surface electromyography sensors. While it was an advantage to use the amputees' prescribed socket and liner because it decreased my overhead cost and time, the disadvantage was that the residual limb sensor site was more susceptible to skin irritation and/or discomfort from concentrated pressure. Due to this increased risk, I was vigilant about monitoring the sensor site and also developed a novel technique for interfacing the sensor with the residual muscle that was effective at minimizing skin irritation and discomfort. In addition to increasing comfort at the sensor site, using custom-designed prosthetic socket-liner systems may have increased the repeatability of my sensor placement. Although custom socket-liner systems may have reduced some of the challenges of the limb-sensor-socket interface, I do not believe that the outcomes of my studies would have been drastically different.

A third general limitation of my study is that my experimental powered prosthesis was fairly cumbersome. The weight of my powered prosthesis was slightly (on average 5 lbs.) heavier compared the subjects' prescribed prosthesis. The experimental prosthesis also had a pendulum effect due to a high concentration of mass at the ankle. I used pneumatic artificial muscles because they have many advantages for human robot system actuation in that they have a high force output to weight ratio and are inherently backdrivable. The disadvantages are that they generate a lot of noise, which some subjects found mildly distracting. They also and they require air hoses, which adds weight to the system. Despite the fact the experimental device was relatively heavy and somewhat loud, it functioned well as a test bed for my continuous proportional myoelectric controller and all of the subjects became accustomed to the device.

The main conclusion to draw from my studies is that there is considerable potential in using residual limb muscles for continuous proportional myoelectric control of powered transtibial prostheses. It is feasible to record robust and reliable electromyography signals from residual muscles at the limb-socket interface by using methods to minimize micro-motions that occur between the electrodes and skin surface. The amputee subjects were able to use continuous proportional myoelectric control to generate biomechanically effective push off at the end of stance phase. Continuous proportional myoelectric control also increased prosthetic embodiment among my users. This control method could enable amputees to perform a wide variety of tasks, adapt to challenging environments, and restore a stronger “sense of self” in future generations of powered lower limb prostheses.

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