ANALYSIS OF UNIQUE MYOELECTRIC CHARACTERISTICS IN LOWER-EXTREMITY MUSCULATURE DURING LOCOMOTIVE STATE TRANSITIONS

by

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A DISSERTATION

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DISSERTATION ABSTRACT

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Lower-extremity amputees face numerous challenges when returning to daily activities. Amongst these challenges is the ability to safely and dynamically transition from one locomotor state to another. Switching between level-ground, ramp, and stair locomotion poses an increased risk as lower-extremity functionality is compromised. Powered prosthetics have been proposed as a solution to this problem. Hypothetically, powered prosthetics would be able to return full functional to the amputated limb. The most common and successful source of information used in algorithms for lower-extremity prosthetics has been electromyography. However, in practice, amputees remain unable to easily actuate the mechanized joints of powered prostheses. Therefore, the current project aimed to identify myoelectric activation differences in lower-extremity musculature during the gait cycles preceding locomotor transition in able-bodied, trans-tibial, and trans-femoral subjects to assist efforts in developing robust classification algorithms for locomotor transitions. Analysis of electromyography was completed to determine if there were periods of activation where classification algorithms could utilize differences in myoelectric activation to appropriately control joint actuation in a subset of eight transitions that included level-ground locomotion and switching to either ramp or stair locomotion and vice versa. Ramp transitions were fundamentally similar to level-ground locomotion and elicited no differences in myoelectric activation. Stair transitions were found to alter muscle activation patterns in able-body and trans-tibial subjects. Trans-femoral subjects differentiated from able-bodied and trans-tibial subjects due to increased recruitment pattern variability. These patterns are distinct and may suggest individual learning patterns within the trans-femoral amputee population. Further investigation of these patterns may be warranted. Findings within able-bodied and trans-tibial subjects suggest common transition based differences within each respective population. Trans-tibial classification algorithms may be developed to utilize this information, using schemes that are focused on important areas during the gait cycle.

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With Aloha for Mom and Dad

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CHAPTER I

INTRODUCTION

1.1 Background and Significance

Lower-limb amputees face numerous challenges in returning to daily activities. Aside from common ailments such as phantom leg pain, a review of secondary amputee pathologies has shown that gait asymmetries result in chronic musculoskeletal ailments of the involved limb, contra-lateral limb, and lower-back (Gailey, 2008). Furthermore, asymmetrical gait causes an increase in the metabolic cost of walking (Bussmann et al., 2008; Caputo and Collins, 2014; Gailey et al., 1994), thus making previously routine activities an extreme burden. The physical strain caused by primary and secondary issues can manifest into significant mental anxiety and depression for amputees (Mckechnie and John, 2014).

It is believed that improving the well-being of, and reducing the incidence of secondary musculoskeletal ailments in, amputees can be accomplished by developing an active prosthesis that mimics the maneuverability of an uninvolved limb. To do this an algorithm, implemented within a microprocessor controller, must be designed to assess sensory feedback. Previous efforts in developing varying types of classification algorithms have used electrooculography, electroencephalography, plantar pressure, inertial measurement units, and electromyography (EMG) (Asghari Oskoei and Hu, 2007; Chen et al., 2015; Chowdhury et al., 2013; Huang et al., 2009; Miller et al., 2012; Tombini et al., 2012; Young et al., 2014). Multisensory algorithms must assess the balance between classification accuracy and computational bandwidth. EMG has emerged as the primary input for classification efforts because it is the neural signal that effects downstream kinematics, such as center of mass, and are assessed via accelerometers.

Neuromuscular action potentials allow muscles to fire in a synergistic pattern to provide the necessary stability and propulsive impulse to maintain balance and forward momentum (Cappellini et al., 2006; Ivanenko et al., 2004; Lacquaniti et al., 2012; Winter and Yack, 1987). However, firing patterns change with differing locomotor states, such as ramp (Franz and Kram, 2012; Lay et al., 2007) and stair (Benedetti et al., 2012; McFadyen and Winter, 1988) ambulation. Previous studies have focused on identifying EMG characteristics in steady state, continuous locomotion. Though beneficial, this provides an incomplete picture of locomotion in daily life.

Transitioning between different forms of dynamic movement is crucial for functioning in daily life. In gait studies, the walk-to-run transition has provided insight into the changes in kinematic (Cappellini et al., 2006; Prilutsky and Gregor, 2001), kinetic (Prilutsky and Gregor, 2001; Sasaki and Neptune, 2006), and EMG (Bartlett and Kram, 2008; Cappellini et al., 2006; Li and Ogden, 2012; Prilutsky and Gregor, 2001; Sasaki and Neptune, 2006) characteristics. For example, kinematic and muscular activation differences may occur for different functions (Cappellini et al., 2006), so timing of pre-transition effect can be established (Li and Ogden, 2012), and changes related to energy efficiency can be determined (Hreljac et al., 2001). In most circumstances, transitions place people in a situation of great instability. In running and cutting, studies have shown that unanticipated cutting tasks yield higher variance in joint angles and muscular activation (Besier et al., 2003; Rand and Ohtsuki, 2000). These tasks require a high level of coordination due to the extreme speed with which these maneuvers are being completed. However, even transitions in the comparatively slower level-ground walking are important to understand.

Previously, muscle activation patterns have been explored for use in musculoskeletal modeling, rehabilitation assessment and treatment, athletic performance, and aspects of algorithm development for mechanized prostheses (Anderson and Pandy, 2003; Cappellini et al., 2006; Enders et al., 2013; Huang et al., 2011; Wakeling and Horn, 2009). Previous attempts at classification algorithms have focused on classifying single state, continuous locomotion (Huang et al., 2011, 2009; Oskoei and Hu, 2008). However, ambulation in everyday environments does not occur in only one terrain. Determining upcoming locomotor transitions (i.e. from level-ground walking to stair ascent or ramp descent to level-ground) is essential for maximizing the utility of an active prosthesis. However, few studies have attempted to identify the characteristic differences in EMG activation during periods of transition between stairs and ramps (Gottschall and Nichols, 2011; Sheehan and Gottschall, 2012, 2011). These studies attempted to identify the gait cycle at which transition begins and how the transition occurs, rapidly or over time, with regard to changing severity of the ensuing state. However, there is yet to be an assessment of the physiological changes in EMG activation profiles from the initial continuous locomotive state, transitional period, and into the second continuous locomotive state across level-ground, ramp, and stair locomotive types.

The outcome of this project may aid the theoretical framework behind locomotor state transitions as it pertains to amputees. Such an understanding would benefit current efforts in classification algorithm development as it provides a much needed bridge to enable amputees using mechanized prostheses to smoothly transition between locomotive states.

1.2 Goals and Specific Aims

Previous attempts at classification algorithms have focused on classifying single state, continuous locomotion. However, ambulation in everyday environments does not occur over only one terrain. Few studies have attempted to identify the characteristic differences in muscle activation during transition between stairs and ramps. Such an understanding would benefit current efforts in classification algorithm development as it provides a much needed bridge to enable mechanized prostheses to smoothly transition between locomotive states.

Specifically, this research aimed to determine whether EMG characteristics in lower-extremity musculature is a suitable source of sensory input for a classification algorithm for mechanized, active prostheses. To address this question, four specific aims were pursued. In most of the study aims, amputees were asked to ambulate across eight different transition types. In Aims 2 and 3, trans-tibial (TT) amputee EMG activation were analyzed in both the involved and uninvolved limbs. In Aim 4, trans-femoral (TF) amputee EMG activation were analyzed in both the involved and uninvolved limbs. In the amputee populations, the involved limb is the limb that sustained the amputation, while the uninvolved limb is the non-amputated, healthy limb. The following aims were pursued in fulfillment of the overall study objective.

Specific Aim 1: Determine whether lower-extremity muscle activation differences between gait cycles occur in able-bodied individuals during locomotive state transitions. It was hypothesized that differences between gait cycles will be observed leading up to and during the gait cycles of transition.

Specific Aim 2: Determine whether lower-extremity muscle activation differences between gait cycles occur in the involved limb of TT amputees during locomotive state transitions. It was hypothesized that differences between gait cycles will be observed leading up to and during the gait cycles of transition.

Specific Aim 3: Determine whether lower-extremity muscle activation differences between gait cycles occur in the uninvolved limb of TT amputees during locomotive state transitions. It was hypothesized that differences between gait cycles will be observed leading up to and during the gait cycles of transition.

Specific Aim 4: Determine whether lower-extremity muscle activation differences between gait cycles occur in the involved and uninvolved limb of TF amputees during locomotive state transitions. It was hypothesized that differences between gait cycles will be observed leading up to and during the gait cycles of transition.

1.3 Summary

Full knowledge of lower-extremity EMG activation differences during able-bodied and amputee terrain transitions would allow for further development of the theoretical construct behind classification algorithms currently being designed for active prostheses. Specifically, identifying the activation differences between gait cycles that a locomotor transition elicits in lower-extremity musculature may help narrow the focus of current classification efforts by highlighting certain gait cycles and/or muscles of import.

One-dimensional, One-Way Statistical Parametric Mapping (SPM) ANOVAs were chosen as the preferred method to determine myoelectric activation differences between pre-transition gait cycles. The SPM statistical paradigm allowed for enhanced identification of neural-mechanical characteristics throughout an entire gait cycle while accounting for Type-I error bias. Since it remains unclear where in the gait cycle myoelectric activation differences may occur, it was important to completely assess influence of the upcoming transitions.

A robust algorithm for transition detection would allow amputees to seamlessly maneuver many obstacles in daily life. The proposed enhancement would be seen as a large step toward bringing a more useable active prosthesis to market. To accomplish this, identification of relevant information is pivotal to reduce bandwidth and energy consumption within an online microprocessor. The general, long-term vision of this research line is to highlight areas of import to 1) improve the information being used in classification algorithms and 2) reduce the utilization of online resources in the microprocessor of active prostheses for both TT and TF amputees.

1.4 Flow of Dissertation

This dissertation is structured using a manuscript style of formatting. The current chapter aimed to provide necessary background information and significance of the overall project. Chapter II aims to provide a general understanding of the methodology used in this dissertation while highlighting the differences between studies. The individual studies contained within Chapters III through VI all use adapted versions of the protocol outlined in Chapter II. These studies are currently in various stages of preparation for submission to peer-review journals. An overall summary of the dissertation findings are then presented in Chapter VII.

Chapter III presents the results of efforts to determine if there are myoelectric activation differences in able-bodied subjects during the three gait cycles leading up to locomotor transition between terrain types.

Chapter IV presents the results of efforts to determine if myoelectric activation differences exist in the involved limb of TT amputees during the three gait cycles leading up to locomotor transition between terrain types.

Chapter V presents the results of efforts to determine if myoelectric activation differences exist in the uninvolved limb of TT amputees during the three gait cycles leading up to locomotor transition between terrain types.

Chapter VI presents the results of efforts to determine if myoelectric activation differences exist in both the involved and uninvolved limb of TF amputees during the three gait cycles leading up to locomotor transitions between terrain types. Though eight transitions were utilized in the previous studies, only six were assessed in this population as the inability to appropriately modulate knee kinetics made two of the transitions unsafe.

A final summary of the dissertation is provided to conclude with key findings from the body of work presented in this dissertation. Additionally, limitations and suggestions for future research are discussed.

CHAPTER II

GENERAL METHODOLOGY

2.1 Recruitment

Able-bodied, unilateral TT amputee, and unilateral TF amputee subjects were recruited for the completion of studies within this dissertation. Inclusion criteria for able-bodied subjects required subjects to have no preexisting lower-extremity musculoskeletal or ligamentous injuries that would inhibit normal gait during level-ground, stair, and ramp locomotion. Inclusion criteria for TT and TF amputees required subjects to have no preexisting lower-extremity musculoskeletal or ligamentous injuries that would inhibit normal gait aside from their amputation. Additionally, amputees were required to be at least one year removed from their last operation related to their amputation. Amputees completed the study using their own passive prosthesis.

2.2 Experimental Protocol

In all studies, surface EMG data were collected from lower-extremity musculature. Passive surface electrodes (Ag/Ag-Cl) were placed on the tibialis anterior (TA), medial gastrocnemius (MG), rectus femoris (RF), vastus lateralis (VL), biceps femoris (BF), gluteus maximus (Gmax), and gluteus medius (Gmed) on the limb of interest using common placement protocols (Delagi et al., 1980) (Figure 2.1). Due to the level of amputation in TF amputees, activation was not acquired from the TA and MG of the involved limb. Muscles located within the prosthetic socket were outfitted with neo-natal electrodes (Figure 2.1; Ambu® BlueSensor NF; Columbia, MD). The neo-natal electrodes are constructed with a lower profile and extended wire that allow for reduced electrode pressure between the residual limb and prosthetic socket; ultimately reducing discomfort and improving signal quality. The TT amputees were outfitted with neo-natal electrodes on the involved limb TA, MG, VL, and, depending on socket sleeve length, BF. The TF amputees were outfitted with neo-natal electrodes on the RF, VL, BF, and, depending on socket sleeve length, Gmax. Muscle belly location of the residual limb musculature was identified by palpating near the typical muscle belly location and asking subjects to activate the appropriate muscle group. The location of maximal muscle mass accumulation was assumed to be the muscle belly.

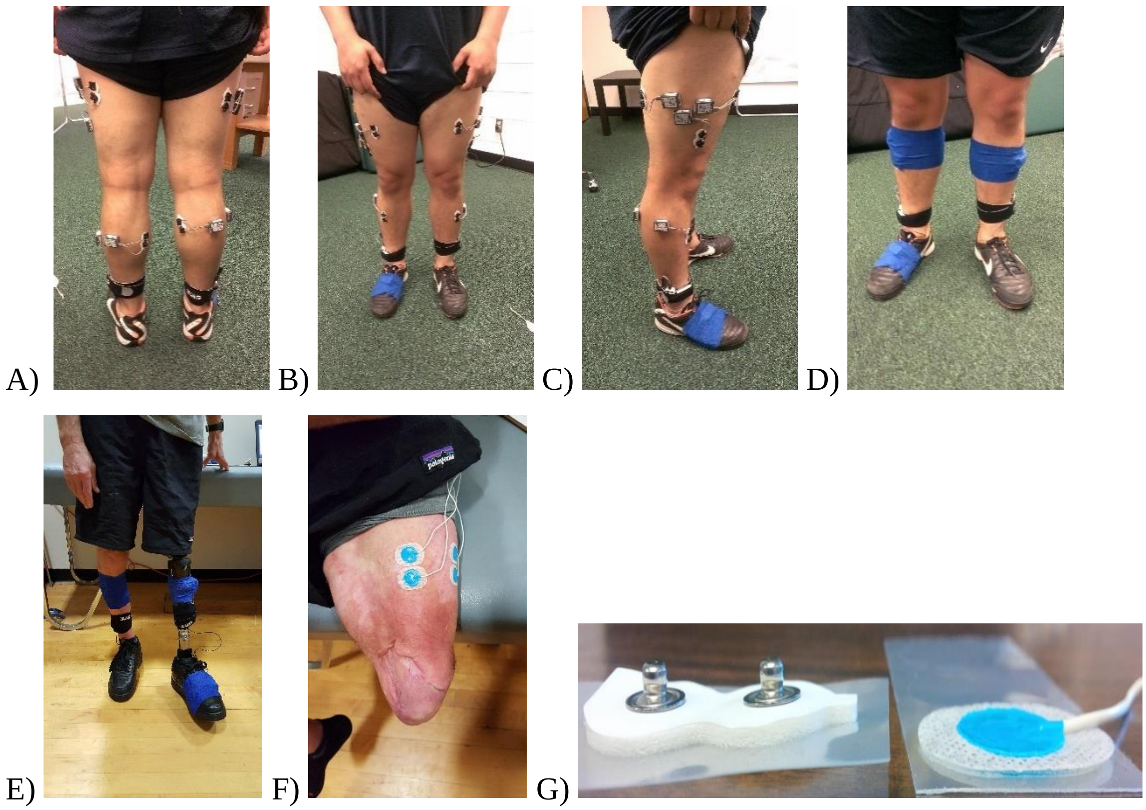


Figure 2.1. Multiple views of EMG electrode, transmitter, and footswitch setup for all studies. *A) Posterior, B) Anterior, and C) Lateral views of subject setup. D) Anterior view of setup with cohesive flexible bandaging. E) Anterior view of TF amputee setup with cohesive flexible bandaging. F) Neo-natal surface electrodes used to collect EMG data from musculature within a prosthetic socket. G) Comparison of depth between original dual passive electrode and neo-natal electrode.*

Local transmitters (Noraxon Telemyo DTS; Scottsdale, AZ) were placed lateral to the collection sites. Cohesive flexible bandaging was used to secure wires and transmitters to reduce motion artifact. Footswitch insoles (Noraxon Telemyo DTS; Scottsdale, AZ) were inserted into each of the subjects' shoes and used to collect the gait events of foot contact and toe off.

2.3 Data Collection

Eight transition types were studied in this dissertation: level ground (LG) to ramp ascent (RA), LG to ramp descent (RD), RA to LG, RD to LG, LG to stair ascent (SA), LG to stair descent (SD), SA to LG, and SD to LG. Trials that utilized the same type of transition, but in the opposite direction, (i.e. LGRA and RDLG) were grouped and alternated to reduce fatigue on the subject. Subjects were asked to begin each trial with a minimum of four gait cycles away from the transition to ensure continuous steady performance of the first locomotion state before transitioning to the second. Transitions were completed on ramps with a grade of 5° and on stairs with a height of 16.5cm and depth of 30.5cm. Subjects in Chapters III, IV, and V were asked to complete a total of 24 successful trials at a self-selected normal walking pace. The TF amputee subjects in Chapter VI were asked to complete a total of 18 successful trials as they were unable to safely complete the LGSA and SDLG transition due to the inability to appropriately modulate involved limb knee kinetics. For each transition, subjects completed three successful trials. A successful trial was defined as transitioning with their involved limb and having completed the trial without any complications. The transitioning limb was defined as the first limb to perform a maneuver whose kinematics were different from the previous state. Therefore, in most conditions, the first limb to land on the ensuing state was considered to be the transitioning limb. However, for SALG and SDLG, the second limb to contact LG was considered the transitioning limb as this limb was judged to perform a kinematically different locomotion.

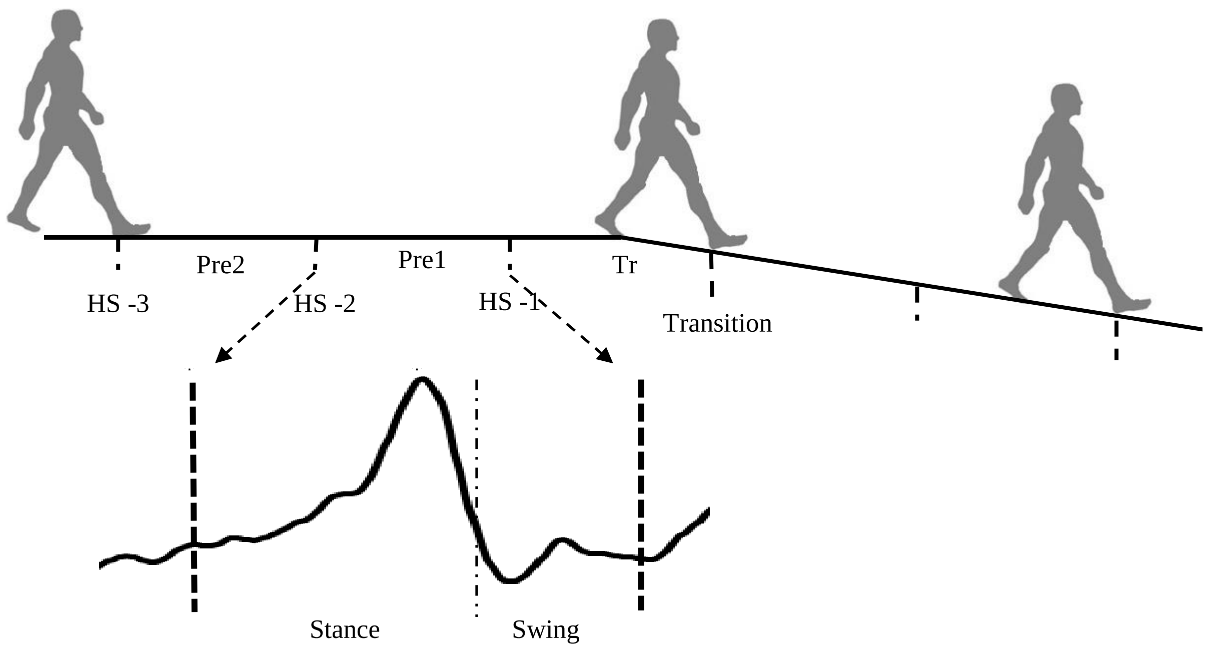


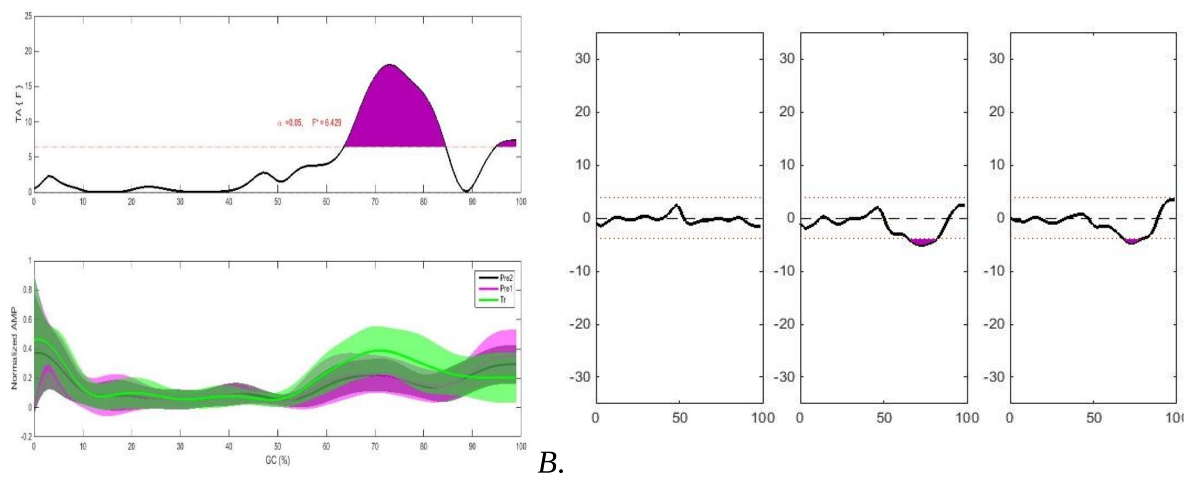
Figure 2.2. Graphical depiction of gait cycle definitions with heel-strike designations. *Gait cycles are defined from heel-strike to heel-strike of the right limb. The enlarged gait cycle shows a description of phases within one representative gait cycle.*

2.4 Data Processing

Three gait cycles per trial were analyzed (Figure 2.2). Cycles were defined from heel-strike to heel-strike of the transitioning limb. The gait cycle beginning three heel-strikes before transition (Pre2), two heel-strikes (Pre1), and one heel-strike (Tr) were analyzed. Each gait cycle was partitioned into three phases: Heel Strike (HS; 0% gait cycle), Stance (ST; first 60% of gait cycle), and Swing (SW; last 40% of gait cycle). The EMG and foot switch data were collected at sampling frequency of 1500 Hz. Gait cycles were ensemble averaged by up-sampling trials, using spline interpolation, to the largest trial length per transition for each subject. The EMG signals were then band-pass filtered (3-500 Hz), full wave rectified, and smoothed using a 4th order Butterworth low-pass filter (6 Hz) to generate a linear envelope that was normalized from 0-100%. Each trial was then amplitude normalized to the maximum peak amplitude of that trial, which was considered to be 100% activation.

2.5 Statistical Analysis

One-dimensional, One-Way Statistical Parametric Mapping (SPM) Analysis of Variance ($\alpha<0.05$) was used in accordance with Pataky (2013) to assess myoelectric activation differences between pre-transitions gait cycles. Figure 2.3A shows representative graphs of the SPM ANOVA results. The bottom graph illustrates the ensemble average of the EMG activation for the three gait cycles preceding transitions for one muscle across all subjects. The top graph illustrates the running f-value of the SPM ANOVA at that point in the gait cycle. The f-critical is designated by the dotted line that runs across the graph. If the f-value exceed the f-critical, the data at that point are considered significantly different.



A. *B.*

Figure 2.3. Representative Graphs of TA SPM ANOVA Results. *Graph A*: *Bottom*-Ensemble averaged EMG activation patterns for Pre2, Pre1, and Tr of a single transition for all subjects. *Top*-Running f-value with f-critical (dotted line). *Graph B*: Post-hoc pairwise comparison of Pre2/Pre1, Pre2/Tr, and Pre1/Tr with running t-statistic and t-critical band (dotted band).

Follow-up, pairwise analyses were also conducted using SPM t-tests, with Bonferroni corrected alpha level ($\alpha<0.017$) to determine differences by gait cycle as subjects approached the transition (Figure 2.3B).

CHAPTER III

PEAK ELECTROMYOGRAPHIC CHARACTERISTICS IN LOWER-LIMB MUSCULATURE PENDING TRANSITION BETWEEN LOCOMOTIVE STATES

3.1 Introduction

Transitioning between different forms of locomotion is crucial for functioning in everyday environments. In gait studies, the walk-to-run transition has provided insight into changing kinematic (Cappellini et al., 2006; Li et al., 1999), kinetic (Prilutsky and Gregor, 2001; Sasaki and Neptune, 2006), and electromyographic (EMG) (Benedetti et al., 2012; Cappellini et al., 2006; Li and Ogden, 2012; Prilutsky and Gregor, 2001; Sasaki and Neptune, 2006) characteristics. Transitions during walking activities can also pose substantial challenges in certain populations (i.e. those with musculoskeletal injury, arthritis, lower limb loss). The characteristics of electromyography (EMG) during locomotor transitions should be studied to bridge the gap between locomotive states.

Neuromuscular action potentials allow muscles to activate in a coordinated sequence to provide the necessary stability and propulsive impulse to maintain balance and forward momentum (Cappellini et al., 2006; Ivanenko et al., 2004; Lacquaniti et al., 2012; Winter and Yack, 1987). However, firing patterns change with differing locomotor states, such as ramp (Franz and Kram, 2012; Lay et al., 2007) and stair (Benedetti et al., 2012; McFadyen and Winter, 1988) ambulation. The previous studies focused on identifying EMG characteristics in steady state, continuous locomotion. Though beneficial, this provides an incomplete picture of locomotion in daily life.

Knowledge about the EMG characteristics governing successful locomotor transitions may aid in the development of robust classification algorithms in active prostheses. Prosthetic control through EMG has been shown to be a viable and accessible source of information (Chowdhury et al., 2013). Current algorithms require repeatable characteristic signatures to reliably classify locomotor types (Graupe et al., 1982). Few studies have identified the characteristic differences in EMG activation during periods of transition between ramps (Gottschall and Nichols, 2011; Sheehan and Gottschall, 2012) or stairs (Joshi et al., 2015; Sheehan and Gottschall, 2011) and level ground.

Previous studies have shown that the transitional period between locomotion states can begin as early as two gait cycles prior to the actual state transition (Gottschall and Nichols, 2011; Sheehan and Gottschall, 2012). The distance at which transitional characteristics were noticed was dependent on the relative difficulty of the ensuing state (Sheehan and Gottschall, 2011). These studies attempted to identify the gait cycle at which transition begins and how the transitions occur, rapidly or over time, with regard to changing severity of the ensuing state (i.e. greater ramp angle). However, little is known regarding the overall characteristic changes in EMG activation from the initial steady state and the transitional period across varying terrains. Identification of unique differences in myoelectric activation patterns may provide valuable insight toward enhancing classification algorithms for powered lower-limb prostheses.

The purpose of this study was to determine differences between pre-transition gait cycles in lower limb muscle activation during transitions between level-ground and ramp/stair locomotor states. It was hypothesized that activation patterns within specific muscles would be observed in gait cycles leading to the transition between locomotor states, for all transitions.

3.2 Methods

Thirteen able-bodied subjects, eleven male (23.5±4.7years; 1.76±0.08m; 76.9±10.0kg) and two female (22.5±0.5years; 1.58±0.51m; 59.8±4.3kg), were recruited for this study. Inclusion criteria required the subjects to have had no preexisting lower body musculoskeletal or ligamentous injuries or pathologies which would inhibit normal gait. All subjects provided written informed consent prior to participation in the IRB-approved protocol.

Surface EMG data were collected from seven muscles of the right limb (Figure 2.1). Passive surface electrodes (Ag/Ag-Cl) were placed on the tibialis anterior (TA), medial gastrocnemius (MG), rectus femoris (RF), vastus lateralis (VL), biceps femoris (BF), gluteus maximus (Gmax), and gluteus medius (Gmed), using common placement protocols (Delagi et al., 1980). Local transmitters (Noraxon Telemyo DTS; Scottsdale, AZ) were placed lateral to the collection sites. Cohesive flexible bandaging was used to secure wires and transmitters to reduce motion artifact. Footswitch insoles (Noraxon Telemyo DTS; Scottsdale, AZ) were inserted into each of the subjects' shoes and used to collect the gait events of foot contact and toe off.

Subjects were asked to complete a total of 24 successful trials at a self-selected normal walking pace. The trials were grouped into eight different transition types: level ground (LG) to ramp ascent (RA), LG to ramp descent (RD), RA to LG, RD to LG, LG to stair ascent (SA), LG to stair descent (SD), SA to LG, and SD to LG. Trials that utilized the same type of transition, but in the opposite direction, (i.e. LGRA and RDLG) were grouped and alternated to reduce fatigue on the subject. Subjects were asked to begin each trial with a minimum of four gait cycles away from the transition to ensure continuous steady performance of the first locomotion state before transitioning to the second. Transitions were completed on ramps with a grade of 5° and on stairs with a height of 16.5cm and depth of 30.5cm. For each transition, subjects completed three successful trials. A successful trial was defined as transitioning with their right limb and having completed the trial without any complications. The transitioning limb was defined as the first limb to perform a maneuver whose kinematics were different from the previous state. Therefore, in most conditions, the first limb to land on the ensuing state was considered to be the transitioning limb. However, for SALG and SDLG, the second limb to contact LG was considered the transitioning limb as this limb was judged to perform a kinematically different locomotion.

Three gait cycles per trial were analyzed in this study (Figure 2.2). Gait cycles were defined from heel-strike to heel-strike of the right limb. The gait cycle beginning three heel-strikes before transition (Pre2), two heel-strikes before transition (Pre1), and one heel-strike before transition (Tr) were analyzed. Gait cycles were partitioned into stance and swing phase at toe-off (~60% of gait cycle). The EMG and foot switch data were collected at sampling frequency of 1500 Hz. Gait cycles were ensemble averaged by up-sampling trials, using spline interpolation, to the largest trial length per transition for each subject. The EMG signals were then band-pass filtered (3-500 Hz), full wave rectified, and smoothed using a 4^{th} order Butterworth low-pass filter (6 Hz) to generate a linear envelope. Each trial was then amplitude normalized to the maximum peak amplitude of that trial, which was considered to be 100% activation.

3.2.1 Statistical Analysis

One-dimensional, One-Way Statistical Parametric Mapping (SPM) Analysis of Variance ($\alpha<0.05$) was used in accordance with Pataky (2013) to assess myoelectric activation differences between pre-transition gait cycles. Figure 2.3A shows representative graphs of the SPM ANOVA results. The bottom graph illustrates the ensemble average of the EMG activation for the three gait cycles preceding transitions for one muscle across all subjects. The top graph illustrates the running f-value of the SPM ANOVA at that point in the gait cycle. The f-critical is designated by the dotted line that runs across the graph. If the f-value exceed the f-critical, the data at that point are considered significantly different.

Follow-up, pairwise analyses were also conducted using SPM t-tests, with Bonferroni corrected alpha level ($\alpha<0.017$) to determine differences by gait cycle as subjects approached the transition (Figure 2.3B).

3.3 Results

Several patterns emerged in the data set. Differences in EMG activation patterns were only observed in the stair transitions, primarily during the swing phase of gait. Additionally, both shank muscles were observed to have significant differences in all stair transitions. The VL and Gmax remained unchanged in all eight transitions. Specific results are presented first by transition, then by muscle, to provide different perspectives by function. Significant findings are presented in Figure 3.1.

3.3.1 By Transition

LGSA: During the LGSA transition, TA, MG, and RF showed significant differences. Most differences were observed during swing phase, the lone exception being a decrease in MG activation during late stance (43-55% of gait cycle). Swing phase activation generally showed an increase with the exception of TA, which decreased in late swing (95-100%).

LGSD: This transition elicited differences in activation of the TA, MG, RF, and Gmed. Similar to the findings in LGSA, significant differences were primarily observed during swing phase, with the exception of MG, in late stance (38-50%). During swing, the TA decreased while the MG, RF, and Gmed all increased activation.

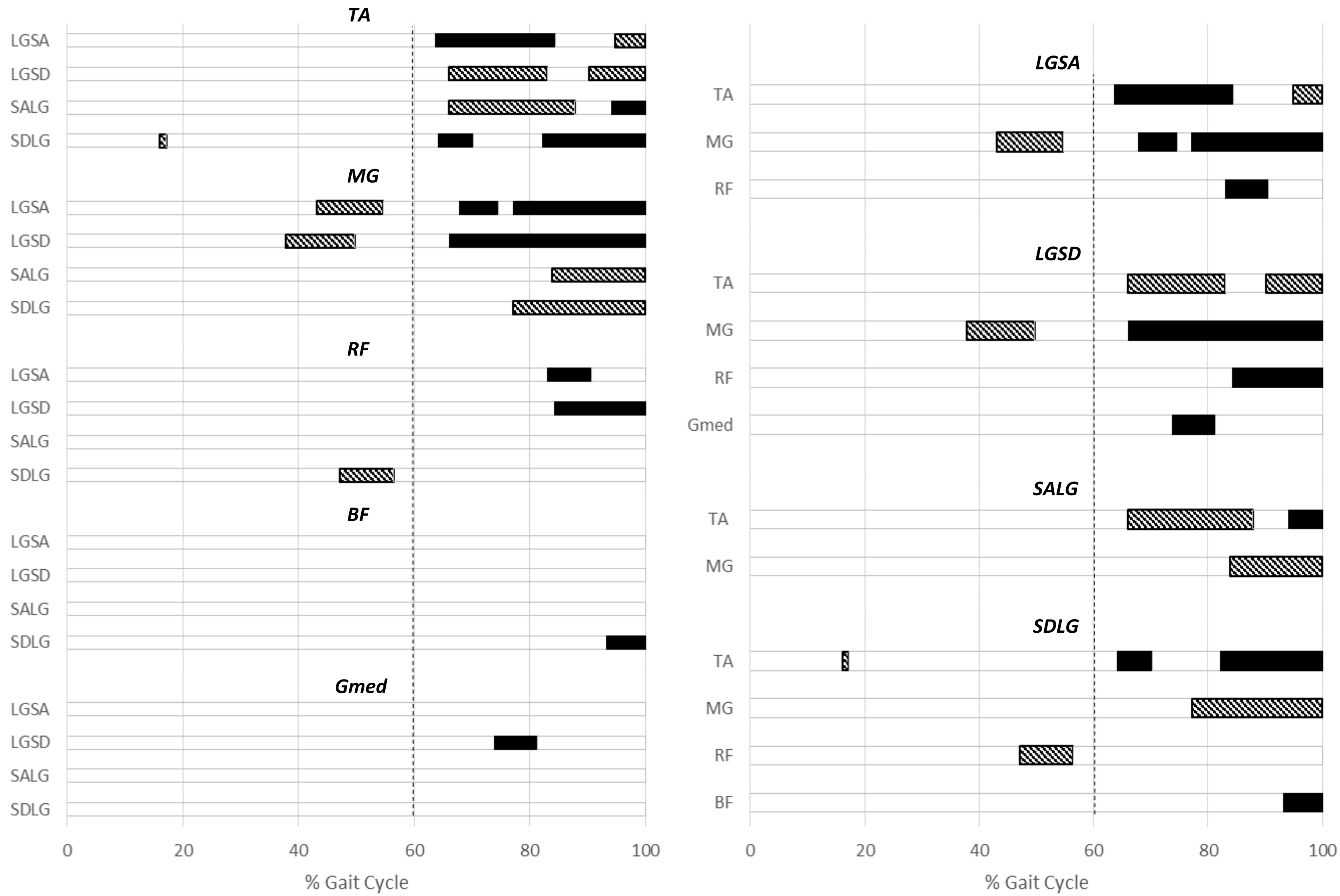


Figure 3.1. Results for Able-Bodied SPM ANOVA from Stair Transitions Grouped by Muscle (Left) and by Transition (Right). Only statistically significant musculature represented. Black bars indicate a significant increase in activation. Striped bars indicate decreased activation.

SALG: Only the shank muscles were affected by this transition. All changes were observed during swing phase. The TA showed decreased activation in early-mid swing (66-88%) then increased activation late (94-100%). The MG was decreased from 82-100%.

SDLG: Four muscles, TA, MG, RF, and BF, were observed to change in this transition. The TA had an early and brief (16-17%) decrease in activation. The RF showed a decrease in activation late in stance (47-56%) and is the only muscle in this transition that did not incur a change during swing. During swing, the TA (64-70%; 82-100%) and BF (93-100%) showed increased activation while the MG (77-100%) decreased.

3.3.2 By Muscle

TA: Differences were primarily observed during the swing phase of the gait cycle. The only exception was a brief decrease during early stance (16-17%) in the SDLG transition. During swing phase, activation differences were observed to occur during early and late swing.

MG: When transitioning from LG to stair locomotion, MG was observed to have decreased activation during late stance and increased activation during most of swing phase. When subjects transitioned from stair locomotion to LG, there was a decrease in activation during late swing phase.

RF: Subjects transitioning from LG to stair locomotion were found to have increased activation in late swing. When subjects moved from SA to LG locomotion, no differences were found. When subjects moved from SD to LG locomotion, however, a decrease in RF activation was observed during late stance (47-56%).

Other Musculature: Two additional muscles were found to have significant changes but only in one transition, respectively. The BF had increased EMG activation during late swing phase of the SDLG transition. The Gmed also had increased activation, but during the mid-swing phase of the LGSD transition.

3.4 Discussion

Previous studies have identified EMG differences in both magnitude and timing during continuous performance of various locomotive types (Franz and Kram, 2012; McIntosh et al., 2006; Winter and Yack, 1987) but have not examined EMG characteristics during the transitional period between different locomotive modes. The current study assessed EMG characteristics for eight locomotor transitions between LG, stairs, and ramps to identify differences in lower-limb muscular activation before transitions.

All stair transitions elicited unique characteristics, while ramp transitions showed no difference in EMG activation. It has been noted previously that stair and ramp locomotion utilizes different kinetics and kinematics (Andriacchi et al., 1980; Franz and Kram, 2012; Lay et al., 2007; McFadyen and Winter, 1988; Sheehan and Gottschall, 2011; Spanjaard et al., 2007; Vallabhajosula et al., 2012) compared to LG locomotion. The current study partially supports those findings, further highlighting the importance of the shank musculature in successfully navigating each transition.

3.4.1 By Transition

LGSA: As subjects walked toward the SA transition, MG decreased activation in late stance phase. This reduction in activation may occur because of the reduced need for a horizontal propulsive mechanism. Rather, the MG is preparing to control the shank for the impending heel-strike on the stair. Additionally, the reduction in activation would reduce co-contractile activation considering the increase in TA activation from early to mid-swing phase. Previous literature supports the notion that this activation pattern would enable the foot to sufficiently dorsiflex for toe clearance over the impending stair (Benedetti et al., 2012; McFadyen and Winter, 1988; Vallabhajosula et al., 2012).

In mid-swing phase, the MG was found to have two portions of higher activation. As the shank is extended over the impending stair, the MG could be acting to provide a knee flexor moment to slow and control the foot position on the ensuing stair (Spanjaard et al., 2007). This idea is further supported by the increased activation of RF during mid-swing. The increased activation would be necessary for two reasons. The first is to act as a hip flexor to lift the thigh, and second to act as an additional knee extensor moment to elevate the shank and foot over the impending stair. The observation of mid to late swing MG activation supports Townsend et al.'s (1978) conclusion that anticipatory plantar flexion is necessary as a damper-like system to absorb energy from foot contact.

LGSD: The activation difference pattern in LGSD is similar to the difference pattern found in LGSA with a few key differences. Similar to the previous transition, a reduction in MG activation during late stance may suggest a reduced need for forward propulsion. A key difference to the LGSA transition is in the TA activation during swing phase. During early swing phase, the TA had decreased activation, as opposed to the increase observed in the LGSA transition. With stair descent, there is reduced concern for sufficient toe clearance (Andriacchi et al., 1980; McFadyen and Winter, 1988; Sinitski et al., 2012). Additionally, reduced TA activation would passively plantar flex the foot making both visual aiming and an anticipatory plantar flexor moment for foot contact easier.

With regard to foot aiming, increased Gmed activation during swing phase may suggest that foot placement, primarily medial/lateral, is important during this portion of the gait cycle. It is known that the Gmed provides pelvic stability during single limb support (Benedetti et al., 2012; Winter et al., 1990), but it also abducts the hip. Increased activation of Gmed during swing may be explained by a need to generate an abduction moment at the hip to provide a mechanism for lateral alignment of the foot on the ensuing stair.

The RF activation mirrors the LGSA with increased mid-swing activation. However, in the LGSD transition, it was also observed that increased activation continues through to late swing phase. This extended increase in activation may suggest anticipatory activation for the impending impulse of an external knee flexor moment from the incurred downward momentum.

SALG: The transition from SA to LG elicited change in the fewest number of muscles. All changes were observed in the two shank muscles. During SA locomotion, toe clearance is critical. However, that need is diminished when transitioning to LG as indicated by the reduced TA activation during early to mid-swing phase. Similarly, in MG, there seems to be a diminished need for an anticipatory plantar flexor moment at the end of swing phase.

It is known that SA locomotion requires increased force output from joint extensors as shown by increased joint extensor moments (Vallabhajosula et al., 2012) and EMG activity (McFadyen and Winter, 1988). Therefore, it could be hypothesized that increased activation during SA locomotion would decrease when transitioning back to LG walking. However, it seems that this shift in the myoelectric activation pattern does not occur within the pre-transition gait cycles.

SDLG: The control of downward momentum is critical in successfully descending stairs. Some of those mechanisms are not required when transitioning to LG locomotion. During late stance phase, the RF was observed to reduce activation. It is known that during this time frame, the RF is providing a knee extensor moment to support body weight and the additional eccentric movement necessitated from going down stairs (McFadyen and Winter, 1988). Transitioning out of stair descent should reduce the need for a knee extensor moment. Current findings of reduced activation during late stance during transition to LG support this notion. Additionally, as subjects moved into swing phase, the TA increased activation while MG decreased. It was previously mentioned that SD locomotion was accomplished with passive plantar flexion and MG control of the shank. While transitioning to LG locomotion, current findings show that those changes are quickly reverted back to LG characteristics.

3.4.2 By Muscle

TA: The important function of the TA is focused on foot positioning. The activation patterns during swing phase seem to correlate well with their transitional counterpart (e.g. LGSA and SALG). The differences in TA activation during continuous LG and SA locomotion were not observed during the transitional gait cycles. In the SDLG transition, it remains unclear if the decreased TA activation from 16-17% of gait cycle is of value.

MG: The activation difference profiles for MG yield two different stories. When subjects were moving from LG to stair locomotion, decreased activation was observed in late stance. This can be attributed to a diminished need for horizontal propulsive force. However, the inverse was not observed when transitioning from stair locomotion to LG. Based on the parameters of this study, it would be expected that the gait cycle immediately after Tr would show increased MG activation. However, the current data do not show strong evidence for this characteristic, pre-transition.

During swing phase, it does appear that the shank controlling mechanism is evident in conjunction with increased anticipatory activation in LG to stair locomotor transitions. However, in stair to LG locomotor transitions, only a decrease in the anticipatory activation was observed. Similar to the findings during stance phase, it would be expected that the difference may be observed in the gait cycle immediately following Tr.

RF: When transitioning from LG to stair locomotion, increased RF activation was observed during late swing. Additionally, the SDLG transition showed a decrease during late stance phase. However, similar findings were not observed in SALG. The lack of difference may suggest that the myoelectric change back to LG locomotion occurs after the Tr gait cycle.

Other Musculature: The BF and Gmed were found to have idiosyncratic differences that do not follow a pattern. Following this study's analysis, it would seem that these differences should not be relied on to identify upcoming locomotor transitions.

3.5 Limitations and Future Work

The current study has a few limitations. First, transitions were observed only at one stair dimension and one ramp inclination angle. The height and depth of the stairs utilized in this study were within Occupational Safety and Health Administration (OSHA) and Building Officials and Code Administrators (BOCA) standards, and the angle of inclination of the ramp was in compliance with the guidelines and specifications for the Americans with Disabilities Act. Differing stair and ramp dimensions may elicit different myoelectric activation characteristics. Second, extraction of other features may possess valuable information in identifying gait transitions between LG and ramp/stair locomotion (Joshi et al., 2015). The current method was used to identify specific areas of locomotion transitions where classification mechanisms may be targeted. Lastly, as the final application of these efforts is in lower limb device control, research must be done to establish whether these findings are similar in the amputee population.

3.6 Conclusion

Ramp transitions did not elicit changes in the musculature observed in the current study. Significant differences were observed, however, in five of the seven lower extremity muscles during stair transitions. Only the TA and MG showed differences in all four stair transitions. The shank musculature seems to provide ample information regarding upcoming transitions, perhaps due to the direct muscular control of the most distal joint. The RF yielded differences in three of the four stair transitions and may be of value in developing classification algorithms for active prostheses. Future research should focus on developing this information in both trans-tibial and trans-femoral amputees.

3.7 Bridge

The study presented in Chapter III was designed to observe what, if any, myoelectric activation differences occurred in the transitioning limb of able-bodied subjects during the three gait cycles leading up to locomotor state transitions. Differences were observed in the able-bodied subjects but remain unknown in TT amputees. Chapter IV explores potential differences in the myoelectric activation pattern of the involved limb in TT amputees. Understanding these differences may begin to influence how classification algorithms for powered lower-extremity prosthetics are designed.

CHAPTER IV

MYOELECTRIC ACTIVATION PATTERN CHANGES IN THE INVOLVED LIMB OF TRANSTIBIAL AMPUTEES DURING LOCOMOTOR STATE TRANSITIONS

4.1 Introduction

Lower-limb amputees face numerous challenges when returning to daily activities. Asymmetrical gait is a common occurrence that propagates into secondary musculoskeletal ailments of the involved limb, contra-lateral limb, and lower back (Gailey, 2008). Active prostheses that mimic the maneuverability of an uninvolved limb have been suggested as a mechanism to reduce the incidence of secondary pathologies. To accomplish this, dynamic control of the active prosthesis is imperative.

Previous studies have had varying levels of success utilizing nervous system activity, in the form of electromyography (EMG), to develop a prosthetic controller (Farmer et al., 2014; Huang et al., 2011, 2014; Ohnishi et al., 2007; Parker et al., 2006). Using EMG, Huang et al. (2011) developed classification algorithms for single-state, continuous locomotion. This would allow amputees to walk with appropriate kinematics on isolated terrain types. Though beneficial, this provides an incomplete picture of locomotion in daily life.

Transitioning between different terrain types is crucial. Previous studies of level-ground (LG) (Winter, 1984), ramp (Lay et al., 2007; McIntosh et al., 2006; Redfern and DiPasquale, 1997), and stair (Andriacchi et al., 1980; McFadyen and Winter, 1988; Riener et al., 2002) locomotion have established that these states require different kinematics. Furthermore, the differences are maintained or exacerbated in trans-tibial (TT) amputees

(Schmalz et al., 2007; Segal et al., 2011). Thus, it is important to focus on the ability of TT amputees with active prostheses to be able to smoothly and safely transition between terrain types.

Few studies have examined myoelectric activation during the actual transition between locomotor states (Gottschall and Nichols, 2011; Sheehan and Gottschall, 2012, 2011). These studies utilized an able-bodied sample. Little is known about muscle activation patterns in TT amputees during the locomotor transitions moments between LG, ramp, and stair terrains. The purpose of this study was to determine whether lower-extremity muscle activation differences between pre-transition gait cycles occur in the involved limb of TT amputees during involved limb locomotive state transitions. It was hypothesized that all transitions would elicit activation differences as the subjects moved closer toward the transition event.

4.2 Methods

Nine unilateral TT amputees (48.8±12.1years; 1.74±0.09m; 86.1±24.7kg) were recruited for this study. Inclusion criteria required the subjects to be at least one year removed from the most recent operation related to the amputation. Average time since amputation was 9.3±9.3 years; ranging from 1-30 years. All subjects provided written informed consent prior to participation in the IRB-approved protocol.

Surface EMG data were collected from seven muscles of the involved limb (Figure 2.1). Passive surface electrodes (Ag/Ag-Cl) were placed on the rectus femoris (RF), vastus lateralis (VL), biceps femoris (BF), gluteus maximus (Gmax), and gluteus medius (Gmed), using common placement protocols (Delagi et al., 1980). Neonatal electrodes (Ambu®

BlueSensor NF; Columbia, MD) were used in lieu of standard surface electrodes on any muscle placement that fell within the prosthetic socket; specifically for the tibialis anterior (TA) and medial gastrocnemius (MG). Muscle belly location for the TA and MG were often compromised as a result of amputation. To identify the location of the muscle belly, palpation of the residual limb was completed while subjects were asked to imagine flexing their toes up (dorsiflexion) and down (plantarflexion). Local transmitters (Noraxon Telemyo DTS; Scottsdale, AZ) were placed lateral to the collection sites. Cohesive flexible bandaging was used to secure wires and transmitters to reduce motion artifact. Footswitch insoles (Noraxon Telemyo DTS; Scottsdale, AZ) were inserted into each of the subjects' shoes and used to collect the gait events of foot contact and toe off.

Subjects were asked to complete a total of 24 successful trials at a self-selected normal walking pace. The trials were grouped into eight different transition types: level ground (LG) to ramp ascent (RA), LG to ramp descent (RD), RA to LG, RD to LG, LG to stair ascent (SA), LG to stair descent (SD), SA to LG, and SD to LG. Trials that utilized the same type of transition, but in the opposite direction, (i.e. LGRA and RDLG) were grouped and alternated to reduce fatigue on the subject. Subjects were asked to begin each trial with a minimum of four gait cycles away from the transition to ensure continuous steady performance of the first locomotion state before transitioning to the second. Transitions were completed on ramps with a grade of 5° and on stairs with a height of 16.5cm and depth of 30.5cm. For each transition, subjects completed three successful trials. A successful trial was defined as transitioning with their involved limb and having completed the trial without any complications. The transitioning limb was defined as the first limb to perform a maneuver whose kinematics were different from the previous state.

Therefore, in most conditions, the first limb to land on the ensuing state was considered to be the transitioning limb. However, for SALG and SDLG, the second limb to contact LG was considered the transitioning limb as this limb was judged to perform a kinematically different locomotion.

Three gait cycles per trial were analyzed in this study (Figure 2.2). Gait cycles were defined from heel-strike to heel-strike of the right limb. The gait cycle beginning three heel-strikes before transition (Pre2), two heel-strikes before transition (Pre1), and one heel-strike before transition (Tr) were analyzed. Gait cycles were partitioned into stance and swing phase at toe-off (~60% of gait cycle). The EMG and foot switch data were collected at sampling frequency of 1500 Hz. Gait cycles were ensemble averaged by up-sampling trials, using spline interpolation, to the largest trial length per transition for each subject. The EMG signals were then band-pass filtered (3-500 Hz), full wave rectified, and smoothed using a 4^{th} order Butterworth low-pass filter (6 Hz) to generate a linear envelope. Each trial was then amplitude normalized to the maximum peak amplitude of that trial, which was considered to be 100% activation.

4.2.1 Statistical Analysis

One-dimensional, One-Way Statistical Parametric Mapping (SPM) Analysis of Variance ($\alpha<0.05$) was used in accordance with Pataky (2013) to assess myoelectric activation differences between pre-transition gait cycles. Figure 2.3A shows representative graphs of the SPM ANOVA results. The bottom graph illustrates the ensemble average of the EMG activation for the three gait cycles preceding transitions for one muscle across all subjects. The top graph illustrates the running f-value of the SPM ANOVA at that point in the gait cycle. The f-critical is designated by the dotted line that runs across the graph. If the f-value exceed the f-critical, the data at that point are considered significantly different.

Follow-up, pairwise analyses were also conducted using SPM t-tests, with Bonferroni corrected alpha level ($\alpha<0.017$) to determine differences by gait cycle as subjects approached the transition (Figure 2.3B).

4.3 Results

Figure 4.1 displays the significant activation differences as determined by SPM ANOVA. Black bars represent significant increases in activation, while striped bars represent decreases in activation during the respective portions of the gait cycle. The left side of the figure groups the significant differences by muscle, while the right side of the figure groups significant differences by stair transition.

Ramp transitions did not elicit a significant change in any of musculature assessed in the current study. Within stair transitions, six muscles showed activation differences with TA being the exception. Individually, these muscles exhibited significant activation differences in no more than two of the four stair transitions. Follow-up pairwise comparison showed that if significant differences were found in SPM ANOVA, the Pre2/Tr and Pre1/Tr comparisons were always significant.

Results will now be presented in a transition by transition and muscle by muscle basis.

4.3.1 By Transition

LGSA: The MG and Gmed were found to have different activation patterns in the LGSA transition. Gmed showed two instances of differing activation. Increased activation

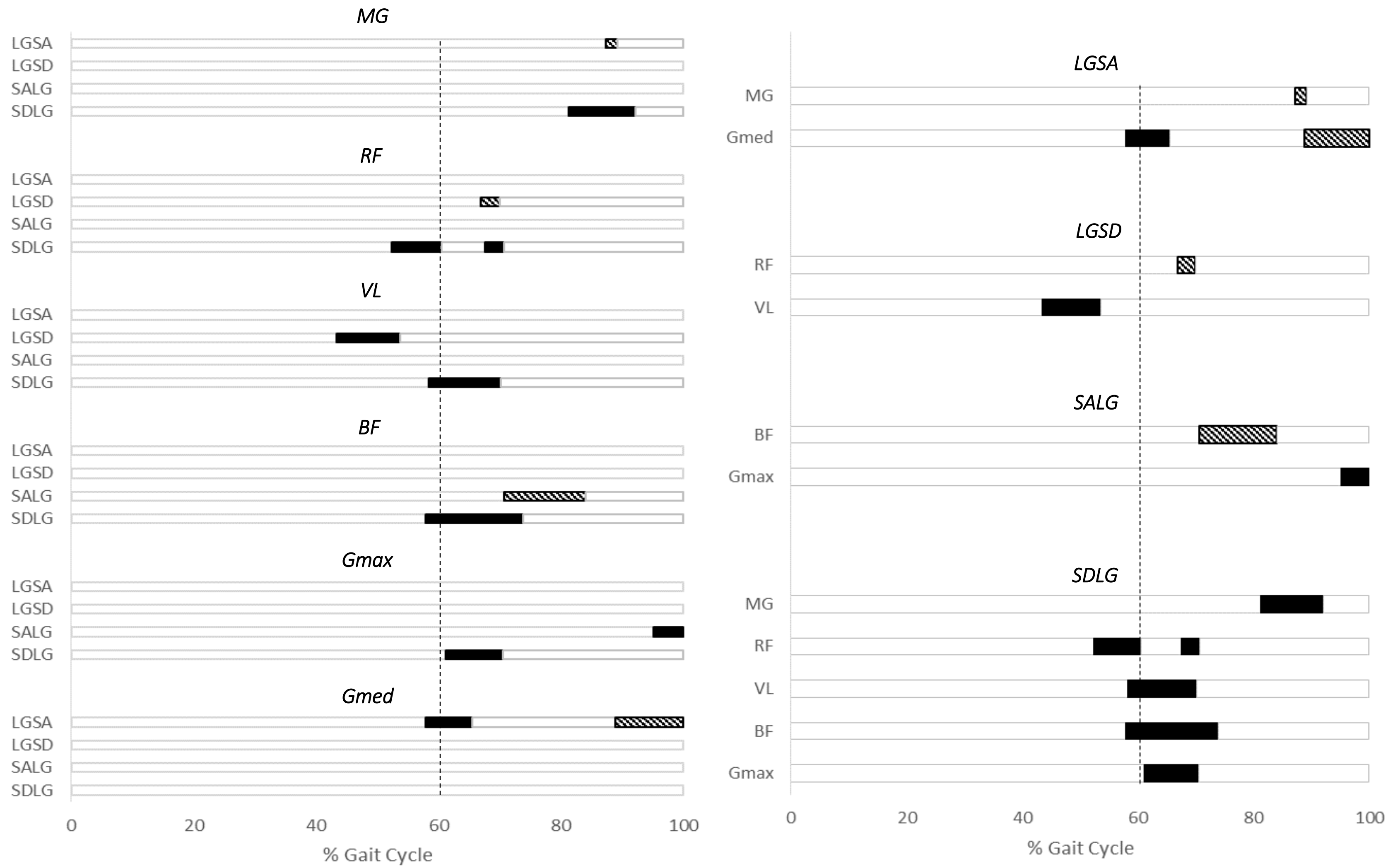


Figure 4.1. Results for TT Amputee Involved Limb SPM ANOVA from Stair Transitions Grouped by Muscle (left) and by Transition (right). Only statistically significant musculature represented. Black bars indicate a significant increase in activation, while striped bars indicate decreased activation. The vertical dashed line represents toe-off between stance and swing phase.

was found in late stance, through early swing (from 58-66% of gait cycle). Late in swing, a decrease in activation was observed from 89-100%. In the MG, a brief decrease in activation was observed from 87-89%.

LGSD: During the LGSD transition RF and VL were observed to have activation differences in swing and stance phases, respectively. The VL had increased activation from 43-54%, while RF showed decreased activation from 67-70%.

SALG: Swing phase showed activation differences in both BF and Gmax during the SALG transition. The BF activation was decreased during swing phase (71-84%) as the subjects moved toward the transition. Late in swing (95-100%), Gmax showed increased activation.

SDLG: The SDLG transition showed altered activation patterns in five of the seven assessed muscles. The differences were found to all be greater levels of activation primarily occurring around toe-off/early swing phase. The RF, VL, BF, and Gmax all showed increased activation primarily from 52-74%. MG showed increased activation from 81-92% of the gait cycle.

4.3.2 By Muscle

MG: Activation patterns of the MG were significantly different in the LGSA and SDLG transitions. These differences were isolated to swing phase. In LGSA, MG showed decreased activation from 87-89%. In SDLG, MG activation increased from 81-92%.

RF: The activation differences found in RF occurred in the reciprocal transitions of LGSD and SDLG during the same phase of gait (67-71%), where there was a significant decrease during the LGSD transition and a significant increase in the SDLG transition. Additionally, the SDLG transition also showed increased activation from 52-61%.

VL: Increased activation in VL occurred in the opposing transitions of LGSD and SDLG but in different phases of gait. In the LGSD transition, a significant increase was observed from 43-54%. In the SDLG transition, the significant increase began around toe-off (58%) and ended in early swing (70%).

BF: Significant differences were observed in the BF as subjects moved from stair locomotion to LG locomotion. Activation decreased during the mid-swing (70-81%) of the SALG transition. During the SDLG transition, increased activation was observed from 58-79%.

Gmax: Similar to the BF, Gmax showed activation differences as subjects transitioned from stair to LG locomotion. These differences were observed in opposing portions of swing phase. The SALG transition elicited increased activation from 95-100%. In the SDLG, increased activation was found during early swing phase (61-71%).

Gmed: The Gmed showed an activation difference in only once transition, LGSA. From 58-66%, activation in the Gmed increased, and in late swing phase (89-100%), Gmed activation decreased.

4.4 Discussion

The aim of this study was to determine if there were myoelectric differences in the musculature of the involved limb in a sample of TT amputees during involved transitions between locomotor states. It was hypothesized that all transitions, to and from ramp and stair locomotion, would yield significant differences. Significant differences were observed in four transitions and six muscles. However, none of the ramp transitions showed myoelectric differences in any of the involved limb musculature. This corroborates findings in able-bodied studies where myoelectric differences were observed, but only at higher grades (Franz and Kram, 2012; Sheehan and Gottschall, 2012). The ramp grade used in this study (5°) would be classified in the low-mid range of the grades assessed in the previous studies. This is the standard set by the Americans with Disabilities Act and represents what would be a common setting in daily life.

The rest of the discussion will focus on the stair transition findings. This section is structured to discuss findings by transition and by muscle.

4.4.1 By Transition

LGSA: The Gmed showed the most prominent difference in this transition. Just before toe-off, Gmed significantly increased activation through early swing. This activation change could be a result of two mechanisms. Amputees are known to have asymmetric gait, which can lead to decreased stability (Adamczyk and Kuo, 2015). Therefore, increased pelvic stability, achieved through Gmed activation would be necessary during this portion of the gait cycle. Additionally, in conjunction with COM sway at toe-off, slight hip abduction of the involved limb would be necessary to maintain linear kinematics during swing.

The late swing deactivation of MG and Gmed begin simultaneously. Since the MG attaches on the medial femoral condyle, brief deactivation could allow subjects to reduce their knee flexor moment to allow the angular momentum of the lower-limb to passively move the limb into appropriate stair kinematics. In combination with MG, the Gmed deactivation may also suggest the utilization of momentum generated earlier in swing. The early swing increase in activation may have slightly abducted the limb, thus the late swing deactivation may correct earlier kinematic differences as the shank progresses toward the ensuing stair.

LGSD: Only quadriceps muscles showed activation differences in the LGSD transition. However, those muscles, RF and VL, acted independently from one another. This may suggest that the dual roles of the RF as a hip flexor and knee extensor were in effect. Late in stance phase, the increased VL activation would suggest that the quadriceps were providing a propulsive knee extensor moment. However, during this portion of stance, the subjects should be extending their hip, requiring RF co-activation.

Early in swing phase, the RF was also observed to briefly decrease activation. The dual role of the RF can also explain this decrease in activation. To transition from LG to SD locomotion, the limb needs to descend toward the ensuing stair, requiring hip and knee extension. Acting eccentrically at the hip, the RF would need to reduce activation to allow passive knee extension. Eccentric hip activation has been reported in LG locomotion, however passive knee extension has not (Lyons et al., 1983).

SALG: Myoelectric changes were observed in the BF and Gmax during SALG but only in swing phase. The deactivation of the BF in mid-swing is interesting because it may be a result of prosthesis compensation. While ascending stairs, the BF is activated to sustain and carry the prosthesis over the ensuing stair (Benedetti et al., 2012). This is necessary due to the fact that the TA no longer controls dorsiflexion to adjust for toe clearance. Thus, amputees will compensate by flexing the knee more to gain sufficient clearance height. However, on the final step before transition, the clearance height is decreased since there is only a single step remaining before LG.

Increased Gmax activation may be a pre-emptive activation for the upcoming gait cycle. In SA locomotion, initial stance requires Gmax to stabilize the hip joint via a hip extensor moment. When transitioning to LG locomotion, the hip extensor moment is increased to accommodate the need to move forward.

SDLG: Most observed changes during SDLG occurred around toe-off. The RF, BF, VL, and Gmax all showed increased activation with temporal onset occurring in that order. The increased activation in multiple muscles suggests that the period between stance and swing is complicated for TT amputees in SDLG transitions. A previous study of ramp locomotion suggested that increased co-activation of musculature was indicative of an upcoming transition (Gottschall and Nichols, 2011). The combined activation effect discovered in this transition may serve as a compensatory mechanism to ensure successful ground clearance by the prosthesis during the initial stages of swing phase. This may be accomplished through a slight delay in hip flexion with Gmax activation, sustained knee flexion with BF and VL co-contraction, and rapid hip flexion from the RF to bring the prosthesis under the body.

The interplay between the RF and Gmax is intriguing. The RF seems to return to a normal level while Gmax activation remains elevated. This may suggest a coordinated effort to maintain an appropriate level of hip joint co-contraction to safely position the thigh. Because the thigh muscles are the most distal, non-compromised musculature in the TT amputees, it may be simpler to manipulate and coordinate these muscles than the most distal, but compromised shank musculature.

4.4.2 By Muscle

Shank: Activation patterns of the shank muscles are non-typical, due to the compromised state of the amputated muscle. Myoelectric activation during transitions was not different in TA, but did change in MG. The LGSA and SDLG transitions elicited a decrease and increase in activation, respectively, during mid-swing.

During mid-swing in able-bodied gait there is usually knee extension with concurrent plantar or dorsiflexion. However, in TT amputees the function of the MG is altered. Though the MG can be activated as a knee flexor it also has the ability to activate and contract to build volume within the socket.

Thigh: Thigh musculature exhibited different activation patterns in three of four transitions, specifically in the SDLG transition. The RF and VL were altered in the LGSD and SDLG transitions. In early swing, RF had opposing myoelectric changes in the reciprocal SD transitions. Additionally, RF was increased pre-toe off in SDLG. This observation is corroborated in previous findings from Benedetti et al. (2012), who reported that the RF had a wide band of activation from mid-stance through mid-swing in level walking. However, when subjects ascended or descended stairs, the activation band narrowed from early to mid-swing. In the current study, the greater activation observed in SDLG appears to be a result of the widening of the RF activation band as subjects transitioned toward LG locomotion.

Myoelectric activation of the BF changed when subjects moved from stair locomotion to LG. Both the SALG and SDLG appear to have been affected by kinematic compensation due to the lack of a plantar-dorsiflexor mechanism in the prosthesis. These findings are similar to those reported by Powers et al. (1997), who concluded that TT amputees alter hip and knee kinematics in continuous stair locomotion. In transitions from stair locomotion, it appears that the myoelectric change associated with the kinematic differences can be observed before the physical transition occurs.

Gluteal: The gluteal musculature is primarily used to provide a hip extensor moment, pelvic stability, and slight hip abduction (Winter, 1983). The current study found that Gmax activation was changed in SALG and SDLG, while Gmed activation was changed in LGSA. The activation differences observed in Gmed, aside from a brief MG deactivation in swing, were the only changes observed as subjects transitioned into SA locomotion. Interestingly, Gmed activation was not observed to change in the inherently more dangerous transitions (e.g. LGSD) where the risk of falling is a serious concern. These differences may be attributed to other sources of sensory input (e.g. vision).

4.5 Limitations and Future Work

This study has some limitations. The gait of TT amputees is varied and depends upon factors such as length of residual limb, amount of gait rehabilitation, secondary musculoskeletal pathologies, and motivation for gait improvement. Locomotor ability was noticeably variable across the subjects in this study. All subjects were able to successfully complete each transition on their own and were not given assistance. If assistance was necessary (i.e. significant handrail use), those trials were not accepted, or evaluated in the analysis. However, individual subject confidence varied and was evident in how they approached the transitions.

Future work should aim to identify whether the uninvolved limb of TT amputees provides insight into the myoelectric mechanisms behind locomotor transitions. The current study's approach could also be used to identify locomotor transition patterns in trans-femoral amputees. A classification algorithm for active trans-femoral prostheses would likely be different from TT algorithms because of the further reduction in viable muscle activation in the involved limb.

The findings in this study provide a snapshot into where classification algorithms could be targeted for enhanced validity and efficiency. The lack of significant differences in ramp transitions begs the question of whether it is important to classify for ramp transitions or whether classification of ramp transitions must utilized some signal source other than EMG.

4.6 Conclusion

Myoelectric activation of lower-extremity musculature was altered in the involved limb of TT amputees when performing transitions involving stair locomotion, but not ramp locomotion. Therefore, EMG as a sensory input for active prostheses may only be useful for stair locomotion. Additional research is needed to determine the viability of having a classifier for stair transitions in combination with alternative strategies for ramp transitions.

4.7 Bridge

In Chapter IV, myoelectric activation differences in the involved limb musculature were identified as TT amputees moved toward upcoming locomotor state transitions. Though differences were identified in the involved limb, it is currently unknown how the uninvolved limb reacts to upcoming transitions. Chapter V will explore potential differences in the uninvolved limb of TT amputees to determine if there are identifiable changes that may aid classification algorithm development.

CHAPTER V

ANALYSIS OF MYOELECTRIC ACTIVATION IN THE UNINVOLVED LIMB OF TRANSTIBIAL AMPUTEES DURING LOCOMOTOR STATETRANSITIONS

5.1 Introduction

By the year 2050, it is conservatively estimated that 3.6 million Americans will be living with an amputation (Ziegler-Graham et al., 2008). The same study estimates that the ratio of lower to upper extremity amputations are 2:1. Thus, it is important for research to focus on providing mechanisms for lower extremity amputees to restore and maintain their quality of life. A primary area of research is to return the amputee's ability to walk in daily life through the use of advanced prosthetics.

Secondary musculoskeletal ailments from asymmetrical gait are a significant concern that can stem from inadequate gait retraining or technology (Gailey, 2008). Passive lower-extremity prostheses perform modestly in returning locomotor capabilities to amputees. However, it is generally understood that passive prosthetics do not return the full functionality of an uninvolved limb. Active lower extremity prostheses aim to return the maneuverability and power generation of uninvolved muscles (Kaufman et al., 2007; van der Linde et al.). The difficulty in using an active prosthesis is in the need to dynamically control the kinetics and kinematics.

Previous studies have used electromyography (EMG) to design prosthetic controllers (Farmer et al., 2014; Huang et al., 2011, 2014; Ohnishi et al., 2007; Parker et al., 2006). The EMG signal is valuable because it can be harnessed locally by the prosthesis and is the neural input that controls muscular activation. Huang et al. (2011) designed a classification algorithm that used EMG to identify bouts of continuous, single-state locomotion. This would allow amputees to continuously walk with the appropriate amount of actuation at each joint. Though beneficial, additional aspects of daily locomotion need to be quantified to provide amputees with a more dynamic set of maneuvers, specifically during transitions between locomotor states.

Locomotor transitions between differing terrain types pose a unique and difficult challenge. Studies have shown that the kinematics of level-ground (LG) (Winter, 1984), ramp (Lay et al., 2007; McIntosh et al., 2006; Redfern and DiPasquale, 1997), and stair (Andriacchi et al., 1980; McFadyen and Winter, 1988; Riener et al., 2002) locomotion are different, and can be challenging for unilateral trans-tibial (TT) amputees (Schmalz et al., 2007; Segal et al., 2011). Reducing the challenge of these transitions is important to reducing injury risk during terrain changes. The development of powered lower-extremity prosthetics provides a way for amputees to regain normal gait patterns. However, it remains unclear how to best control the powered prosthetic systems during terrain changes.

Few studies have studied EMG activation patterns while walking across terrain changes (Gottschall and Nichols, 2011; Sheehan and Gottschall, 2012, 2011). Further, the previous studies were completed in able-bodied individuals and not amputees. It remains unclear whether the EMG activation patterns are generalizable to the amputee population. Amputee residual limb EMG data are variable due to the compromised state of the musculature (Ivanenko et al., 2013). Uninvolved limb EMG patterns may deliver more consistent and less variable EMG signals where characteristics influenced by upcoming terrain changes are more identifiable. The purpose of this study was to determine whether lower-extremity muscle activation differences between pre-transition gait cycles occur in the uninvolved limb of TT amputees during involved limb locomotive state transitions. It was hypothesized that all transitions would elicit muscle activation differences as the subjects moved toward the transition.

5.2 Methods

Nine unilateral TT amputees (48.8±12.1years; 1.74±0.09m; 86.1±24.7kg) were recruited for this study. Inclusion criteria required the subjects to be at least one year removed from the most recent operation related to the amputation. Average time since amputation was 9.3±9.3 months; ranging from 1-30 years. All subjects provided written informed consent prior to participation in the IRB-approved protocol.

Surface EMG data were collected from seven muscles of the uninvolved limb (Figure 2.1). Passive surface electrodes (Ag/Ag-Cl) were placed on the tibialis anterior (TA), medial gastrocnemius (MG), rectus femoris (RF), vastus lateralis (VL), biceps femoris (BF), gluteus maximus (Gmax), and gluteus medius (Gmed), using common placement protocols (Delagi et al., 1980). Local transmitters (Noraxon Telemyo DTS; Scottsdale, AZ) were placed lateral to the collection sites. Cohesive flexible bandaging was used to secure wires and transmitters to reduce motion artifact. Footswitch insoles (Noraxon Telemyo DTS; Scottsdale, AZ) were inserted into each of the subjects' shoes and used to collect the gait events of foot contact and toe off.

Subjects were asked to complete a total of 24 successful trials at a self-selected normal walking pace. The trials were grouped into eight different transition types: LG to ramp ascent (RA), LG to ramp descent (RD), RA to LG, RD to LG, LG to stair ascent (SA), LG to stair descent (SD), SA to LG, and SD to LG. Trials that utilized the same type of transition, but in the opposite direction, (e.g. LGRA and RDLG) were grouped and alternated to reduce fatigue on the subject. Subjects were asked to begin each trial with a minimum of four gait cycles away from the transition to ensure continuous locomotion of the first gait type before transitioning to the second. Transitions were completed on ramps with a grade of 5° and on stairs with a height of 16.5cm and depth of 30.5cm. Subjects completed three successful trials per transition. A successful trial was defined as transitioning with the subject’s involved limb and having completed the trial without any complications. The transitioning limb was defined as the first limb to perform a maneuver whose kinematics were different from the previous state. Therefore, in most conditions, the first limb to land on the ensuing state was considered to be the transitioning limb. However, for SALG and SDLG, the second limb to contact LG was considered the transitioning limb as this limb performs the kinematically different locomotion.

Three gait cycles per trial were analyzed in this study (Figure 2.2). Gait cycles were defined from heel-strike to heel-strike of the involved limb. The gait cycle beginning three heel-strikes before transition (Pre2), two heel-strikes before transition (Pre1), and one heel-strike before transition (Tr) were analyzed. Gait cycles were partitioned into stance and swing phase at toe-off (60% of gait cycle). The EMG and foot switch data were collected at sampling frequency of 1500 Hz. Gait cycles were ensemble averaged by up-sampling trials, using spline interpolation, to the largest trial length per transition for each subject and time normalized from 0-100%. The EMG signals were then band-pass filtered (3-500 Hz), full wave rectified, and smoothed using a 4^{th} order Butterworth low-pass filter (6 Hz) to generate a linear envelope. Each trial was then amplitude normalized to the maximum peak amplitude of that trial.

5.2.1 Statistical Analysis

One-dimensional, One-Way Statistical Parametric Mapping (SPM) Analysis of Variance ($\alpha<0.05$) was used in accordance with previous literature (Pataky et al., 2013; Robinson et al., 2015) to assess differences between pre-transition gait cycles. An SPM ANOVA was completed for each muscle in eight transitions. Figure 2.3A shows representative graphs of the SPM ANOVA results. The bottom graph illustrates the ensemble average of the EMG activation for the three gait cycles preceding transitions for one muscle across all subjects. The top graph illustrates the running f-value of the SPM ANOVA at that point in the gait cycle. The f-critical is designated by the dotted line that runs across the graph. If the f-value exceeds the f-critical, the data at that point are considered significantly different.

Follow-up, pairwise analyses were also conducted using SPM t-tests, with Bonferroni corrected alpha level ($\alpha<0.017$) to determine differences by gait cycle as subjects approached the transition (Figure 2.3B).

5.3 Results

Figure 5.1 displays the significant activation differences as determined by SPM ANOVA. Ramp transitions did not elicit a significant change in any muscles assessed in the current study. However, all muscles were found to have activation differences in at least one stair transition. The VL was observed to have a difference in all four stair transitions. All temporal references are with respect to involved limb gait cycles.

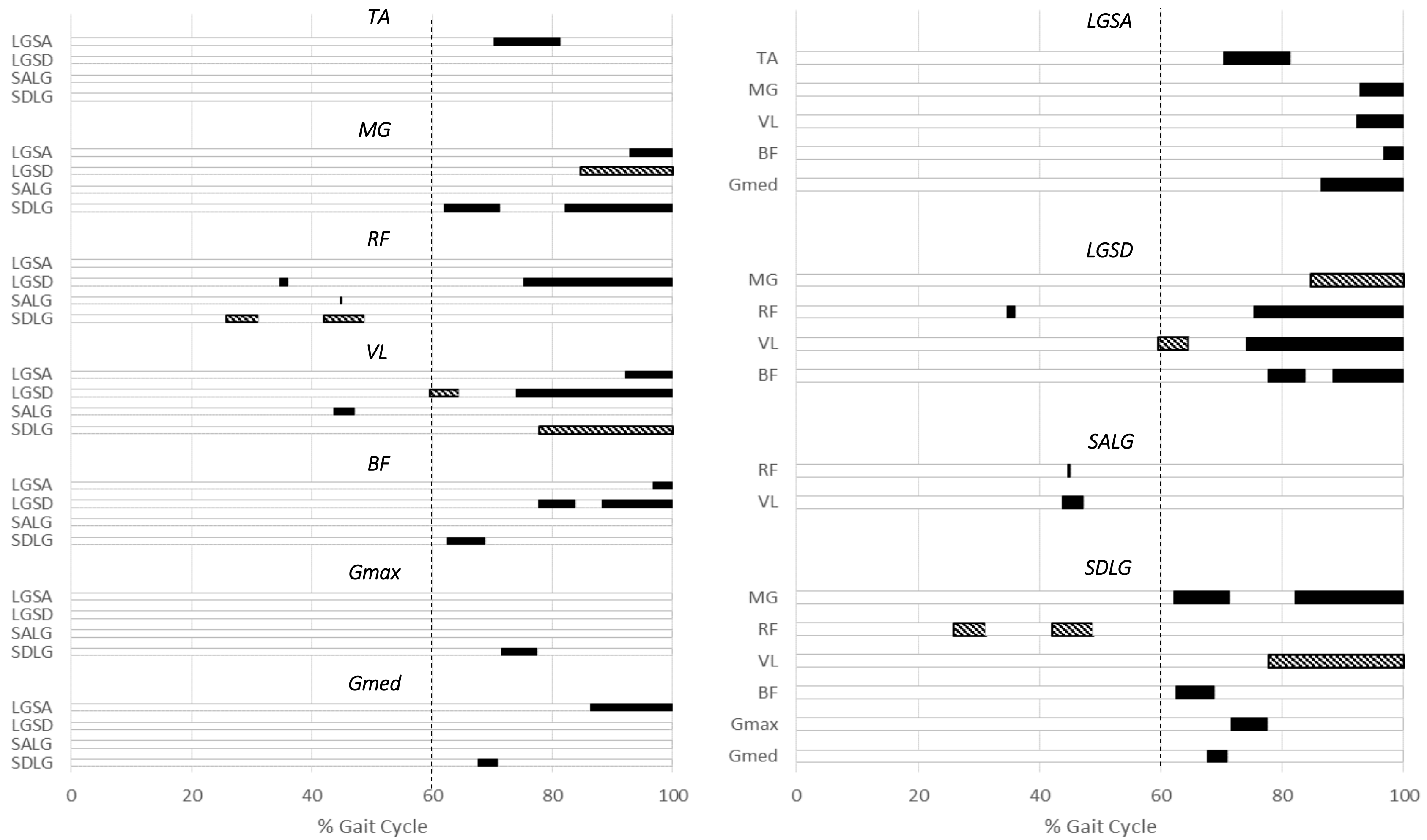


Figure 5.1. Results for TT Amputee Uninvolved Limb SPM ANOVA from Stair Transitions Grouped by Muscle (Left) and by Transition (Right). Only statistically significant musculature represented. Black bars indicate a significant increase in activation, while striped bars indicate decreased activation. The vertical dashed line represents toe-off between stance and swing phase.

Findings are first presented with reference to each transition, and then with reference to each muscle.

5.3.1 By Transition

LGSA: During the LGSA transition, activation of the uninvolved limb TA, MG, VL, BF, and Gmed were all increased during the involved limb swing phase of the transition gait cycle. The TA differences occurred earliest, from 70-81% of the gait cycle. Sequentially, Gmed (86%), VL (92%), MG (93%), and BF (97%) showed increased activation through the end of the transition gait cycle.

LGSD: The uninvolved limb thigh musculature, along with MG, were observed to have characteristic differences during the LGSD transition. Thigh musculature all showed similar increased activation from approximately 75-100%. Individually, RF activation was increased from 35-37% of the gait cycle, while VL deactivated near toe-off (60-65%). From 85-100%, there was relative deactivation of the MG.

SALG: During the SALG, both quadriceps muscles were found to have short periods of increased activation during stance phase (RF: 45%; VL: 43-47%).

SDLG: Six muscles (all but TA) were found to have characteristic differences in activation during SDLG. Deactivation was observed in the RF (26-31%, 42-49%) and VL (78-100%). Increased activation was observed in the MG (62-71%, 82-100%), BF (63-69%), Gmax (72-78%), and Gmed (68-71%). The only change observed during stance phase was in the RF.

5.3.2 By Muscle:

TA: The LGSA transition was the only condition in which a change in TA activation (70-81%) was observed.

MG: Activation differences were observed in the MG during three different transitions and all during swing phase. The LGSA and SDLG transitions elicited increased activation patterns, while LGSD elicited deactivation pattern in MG. The increased activation found in SDLG occurred in two periods; 62-71%, and 82-100%.

RF: Short windows of activation differences were observed during involved limb stance phase in the LGSD, SALG, and SDLG transitions. The SDLG transition was found to have two periods of deactivation from 26-31% and 42-49%. The LGSD transition also elicited a period of increased activation from mid to late swing phase.

VL: The VL was the only muscle to show activation differences in all four stair transitions. Differences were observed during involved limb swing phase in LGSA (92-100%), LGSD (85-100%), and SDLG (62-71%; 82-100%) transitions. The LGSD transition resulted in two periods of VL activation differences with early swing deactivation and mid to late swing increase in activation.

BF: Differences observed in the BF were found to show only increases in myoelectric activation, occurring in the LGSA (97-100%), LGSD (78-84%; 88-100%), and SDLG (63-69%) transitions. All differences were observed during involved limb swing phase.

Gmax: A single increase in activation was observed for Gmax in the SDLG transition during involved limb swing phase (72-78%).

Gmed: Activation differences of the Gmed were observed in the LGSA and SDLG transitions. These were both increases in activation during swing phase; 86-100% and 68-71%, respectively.

5.3.3 Lead-Up Effect:

Results of the follow-up pairwise comparisons were consistent in all ANOVA SPM analyses. If the SPM results were found to be significant, differences were always found in the Pre2/Tr and Pre1/Tr comparisons. If the SPM results were not significant, no follow-up comparisons were found to be significant.

5.4 Discussion

The aim of this study was to determine if there were myoelectric differences in the musculature of the uninvolved limb of TT amputees during involved limb transitions between locomotor states. Further, the aim was to identify these changes within the time perspective of involved limb transitions. It was hypothesized that all ramp and stair transitions would yield significant differences. However, all four ramp transitions showed no activation differences in any of the muscles of the uninvolved limb. This may be partially explained by previous able-bodied studies where observable myoelectric differences were only found at higher ramp inclines (Franz and Kram, 2012; Sheehan and Gottschall, 2012). The current study used a 5° incline that would be classified in the low to middle ranges of the previous studies. In perspective, the 5° incline is the standard inclination angle set forth by the Americans with Disabilities Act and would represent a common setting in daily life.

Similarly, the stairs that were used in this study were within standard dimensions set forth by the Occupational Safety and Health Administration (OHSA) and the Building Officials and Code Administrators (BOCA). However, significant differences in myoelectric activation were indeed identified in stair transitions.

5.4.1 Swing Phase

A majority of the differences identified were during the involved limb swing phase portion of the gait cycle. In typical level ground locomotion, ipsilateral limb swing phase represents the period of time where the contralateral limb is in stance phase. In TT amputee gait, biomechanical compensatory mechanisms of the uninvolved limb typically occur to comfortably maneuver the involved limb through swing phase.

This finding may suggest that during the period of involved limb swing phase, the uninvolved limb provides valuable transition-specific information. Further, the lack of myoelectric differences during stance phase suggests that classification algorithms do not need to consider uninvolved limb EMG activity during this time period (Nolan et al., 2003).

5.4.2 Thigh Musculature

Throughout the gait cycle, the thigh musculature provide internal moments that stabilize the knee and hip joints during stance phase, accelerate the limb through swing phase, and adjust limb kinematics to avoid collisions with physical objects (e.g. stairs) (Andriacchi et al., 1980; Benedetti et al., 2012). In unilateral amputees, the appropriate motor control of the uninvolved limb, the only completely uninvolved lower-extremity limb, is paramount. In the current study, the VL showed activation differences in all four stair transitions. Additionally, RF and BF activation levels were found to be different in three stair transitions. Beyond these three muscles, only MG activation was changed in more than two transitions.

This finding highlights the importance of the thigh musculature, not only as it pertains to kinematic and kinetic functions, but toward defining locomotion transition classification mechanisms. Myoelectric changes were observed in the TA, Gmax, and

Gmed, however, the value of incorporating another source of information versus removing it for system efficiency is unclear. With computational and battery efficiency being a primary concern in the development of powered prosthetics (Joshi et al., 2016), the reduction of unnecessary equipment and/or information may be just as useful as knowing which signals change the most.

5.4.3 Stair Ascent to Level Ground

The SALG transition yielded the least amount of change in myoelectric activation. Furthermore, the changes that were observed were for brief periods during the gait cycle. In contrast to previous findings about the data, the changes identified were only during stance phase of the involved limb. Though the lack of identifiable differences could be a detriment, the phase difference in where observed myoelectric changes are occurring could be strength. Nonetheless, it remains unclear how the lack of identifiable differences would alter classification algorithms.

5.4.5 Lead-Up Effect

Differences found in the SPM ANOVAs are primarily attributed to the differences observed in the Tr gait cycle. Sheehan & Gottschall(Sheehan and Gottschall, 2011) previously concluded that activation differences during stair transitions are instantaneous and occur very close to the stair. Follow-up pairwise comparisons corroborate this finding with no differences found in the Pre2/Pre1 gait cycle comparison.

5.5 Limitations and Future Work

The use of amputees with passive lower-extremity prosthetics is a limitation in this study. Many of the myoelectric activation patterns will likely translate between amputees who use a passive versus powered prosthetic. Nonetheless, future work should attempt to identify if there are any myoelectric differences in lower-extremity musculature when switching from a passive to a powered prosthetic. Future work should also quantify the computational value of including uninvolved limb EMG into a classification algorithm. Though the current study has identified areas where differences are present, advanced algorithms may be able to use uninvolved limb EMG as a primary or secondary source of information.

5.6 Conclusion

Ramp transitions did not elicit any discernable myoelectric differences in the muscles of the uninvolved limb. In stair transitions, it was found that the thigh musculature provided most of the differences observed. Further, the activation of the VL was changed in all four stair transitions. Swing phase of the involved limb, especially mid to late swing, elicited the greatest amount of activation differences in the musculature of the uninvolved limb.

5.7 Bridge

Chapters IV and V presented the findings of myoelectric activation differences in both the involved and uninvolved limbs of TT amputees as locomotor state transitions are approached. These findings provide valuable information regarding classification algorithms for TT amputee prostheses. However, prostheses used by TF amputees may necessitate an original algorithm that accounts for anatomical and muscle recruitment differences. Chapter VI will investigate potential myoelectric activation changes in both the involved and uninvolved limb musculature of TF amputees.

CHAPTER VI

INVOLVED AND UNINVOLVED LIMB MYOELECTRIC ACTIVATION DIFFERENCES IN TRANSFEMORAL AMPUTEES DURING LOCOMOTOR STATE TRANSITIONS

6.1 Introduction

According to current amputation rates, by the year 2050, amputees will account for approximately 1% of the total American population (Ziegler-Graham et al., 2008). The same study also estimates the prevalence of lower-extremity amputees to be double the population of upper-extremity amputees. In lower-extremity amputees, secondary musculoskeletal ailments are common and are associated with poor gait characteristics during locomotion (Gailey, 2008). Poor gait characteristics are attributed to passive prostheses that do not provide adequate knee and/or ankle actuation during specific phases of gait.

Active lower-extremity prostheses aim to return the original maneuverability and power generation provided by uninvolved musculature (Au et al., 2007; Caputo and Collins, 2014; Sup et al., 2008). Though active prostheses are considered to be beneficial for amputees, it remains challenging to dynamically control the actuation of the knee and ankle mechanisms. Previous studies have aimed to develop a robust controller for active prostheses (Huang et al., 2014; Ohnishi et al., 2007; Oskoei and Hu, 2008). Recently, studies have used electromyography (EMG) as the primary component for powered prosthetic controllers (Farmer et al., 2014; Huang et al., 2011, 2009; Joshi et al., 2015; Ohnishi et al., 2007; Parker et al., 2006).

The EMG signal is believed to be a valuable sensory input because it is a non-invasive neural signal that is relatively easy to acquire. Previous studies have been able to use EMG data to classify various bouts of continuous, single-state locomotion (Huang et al., 2011, 2009). Classification algorithms are important because they provide decision assistance to powered prosthetics in how to actuate the knee and ankle mechanisms. Single-state locomotion is important to control efficiently, but it does not account for the majority of movements that occur in daily life.

Switching between differing locomotion modes can pose a unique challenge when transitioning from one terrain type to another. Level-ground (LG) (Winter, 1984), ramp (Lay et al., 2007; McIntosh et al., 2006; Redfern and DiPasquale, 1997), and stair (Andriacchi et al., 1980; McFadyen and Winter, 1988; Riener et al., 2002) locomotion are known to be different mechanical challenges, requiring different kinematic solutions. These differences can be exacerbated in unilateral trans-femoral (TF) amputees (Bae et al., 2007; Boonstra et al., 1994; Hobara et al., 2013; Jaegers et al., 1995; Kaufman et al., 2007; Schmalz et al., 2007; Segal et al., 2006). The use of powered prosthetics in TF amputees may allow amputees to overcome these challenges by providing the appropriate knee and ankle actuation. However, it remains unclear how to best control a powered prosthesis while transitioning between differing terrains.

Few studies have investigated potential EMG activation pattern differences during terrain transitions (Gottschall and Nichols, 2011; Sheehan and Gottschall, 2012, 2011). The previous studies concluded that pre-ramp transition activation differences were a function of grade severity (i.e. steeper grade change yielded earlier activation) (Sheehan and Gottschall, 2012), while pre-stair transition differences were instantaneous (i.e. within the final gait cycle) (Sheehan and Gottschall, 2011). Appreciable pre-transition muscle activation (EMG) differences seem to be obtainable in able-bodied individuals. There is a need to assess muscle activation differences in TF amputees to determine the validity in using EMG as a sensory input for powered prosthetics. The purpose of this study was to determine whether lower-extremity muscle activation differences between pre-transition gait cycles occur in both the involved and uninvolved limb of TF amputees during involved limb locomotive state transitions. It was hypothesized that all transitions would elicit muscle activation differences as the subjects moved toward the transition.

6.2 Methods

Five unilateral TF amputees (50.8±13.5years; 1.70±0.05m; 78.7±12.1kg) were recruited for this study. Inclusion criteria required the subjects to be at least one year removed from the most recent operation related to the amputation. Average time since amputation was 17.1±12.8 years; ranging from 3.0-32.9 years. All subjects provided written informed consent prior to participation in the IRB-approved protocol.

Surface EMG data were collected from five muscles of the involved limb and seven muscles of the uninvolved limb (Figure 2.1). Low-profile, neo-natal surface electrodes (Ambu® BlueSensor NF; Columbia, MD) were placed on the residual limb rectus femoris (RF), vastus lateralis (VL), and biceps femoris (BF). Muscle belly location was identified by palpation near typical muscle belly locations while asking subjects to activate the appropriate muscle group. Standard passive surface electrodes (Ag/Ag-Cl) for the involved limb were placed on the gluteus maximus (Gmax) and gluteus medius (Gmed), using common placement protocols (Delagi et al., 1980). Standard passive surface electrodes of the uninvolved limb were placed on aforementioned muscles as well as the tibialis anterior (TA) and medial gastrocnemius (MG). Local transmitters (Noraxon Telemyo DTS; Scottsdale, AZ) were placed lateral to the collection sites. Cohesive flexible bandaging was used to secure wires and transmitters to reduce motion artifact. Footswitch insoles (Noraxon Telemyo DTS; Scottsdale, AZ) were inserted into each of the subjects' shoes and used to collect the gait events of foot contact and toe off.

Subjects were asked to complete a total of 18 successful trials at a self-selected normal walking pace. The trials were grouped into six different transition types: LG to ramp ascent (RA), LG to ramp descent (RD), RA to LG, RD to LG, LG to stair descent (SD), and SA to LG. The LGSA and SDLG transitions were considered but omitted due to safety concerns regarding TF amputees using their involved limb for transition without the ability to modulate knee kinetics. Trials that utilized the same type of transition, but in the opposite direction, (e.g. LGRA and RDLG) were grouped and alternated to reduce fatigue on the subject. Subjects were asked to begin each trial with a minimum of five gait cycles away from the transition to ensure continuous locomotion of the first gait type before transitioning to the second. Transitions were completed on ramps with a grade of 5° and on stairs with a height of 16.5cm and depth of 30.5cm. For each transition, subjects completed three successful trials. A successful trial was defined as transitioning with the subject's involved limb and having completed the trial without any complications. The transitioning limb was defined as the first limb to perform a maneuver whose kinematics were different from the previous state. Therefore, in most conditions, the first limb to land on the ensuing state was considered to be the transitioning limb. However, for SALG and

SDLG, the second limb to contact LG was considered the transitioning limb as this limb performs the kinematically different locomotion.

Three gait cycles per trial were analyzed in this study (Figure 2.2). Gait cycles were defined from heel-strike to heel-strike of the involved limb. Involved limb gait cycles were used as the time reference to assess both the involved and uninvolved limb EMG. The gait cycle beginning three heel-strikes before transition (Pre2), two heel-strikes before transition (Pre1), and one heel-strike before transition (Tr) were analyzed. Gait cycles were partitioned into stance and swing phase at toe-off (60% of gait cycle). The EMG and foot switch data were collected at sampling frequency of 1500 Hz. Gait cycles were ensemble averaged by up-sampling trials, using spline interpolation, to the largest trial length per transition for each subject and time normalized from 0-100%. The EMG signals were then band-pass filtered (3-500 Hz), full wave rectified, and smoothed using a 4th order Butterworth low-pass filter (6 Hz) to generate a linear envelope. Each trial was then amplitude normalized to the maximum peak amplitude of that trial.

6.2.1 Statistical Analysis

One-dimensional, One-Way Statistical Parametric Mapping (SPM) Analysis of Variance ($\alpha<0.05$) was used in accordance with previous literature (Pataky et al., 2013; Robinson et al., 2015) to assess difference between pre-transition gait cycles. SPM ANOVAs were completed to assess each muscle in all six transitions. Figure 2.3A shows representative graphs of the SPM ANOVA results. The bottom graph illustrates the ensemble average of the EMG activation for the three gait cycles preceding transitions for one muscle across all subjects. The top graph illustrates the running f-value of the SPM ANOVA at that point in the gait cycle. The f-critical is designated by the dotted line that runs across the graph. If the f-value exceeds the f-critical, the data at that point are considered significantly different.

Follow-up, pairwise analyses were also conducted using SPM t-tests, with Bonferroni corrected alpha level (α<0.017) to determine differences by gait cycle as subjects approached the transition (Figure 2.3B).

6.3 Results

One-dimensional SPM ANOVA analyses of involved limb musculature for the gait cycles leading up to transition yielded no significant differences in any of the musculature in all six transitions. One significant difference was identified in uninvolved limb activation during the SALG transition in the VL (Figure 6.1).

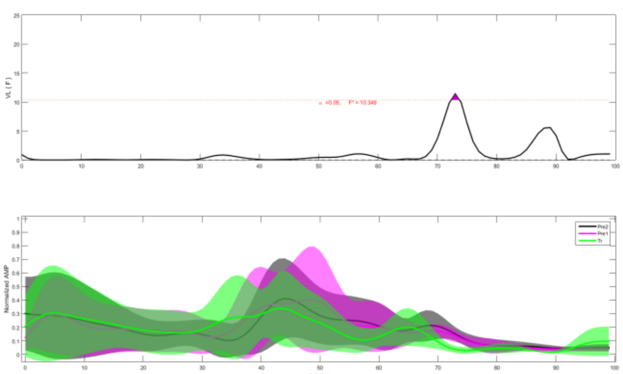


Figure 6.1 One-Dimensional SPM ANOVA Result for Uninvolved Limb VL During the SALG Transition Highlighting the Only Significant Difference Observed.

Table 1 shows subject-specific SPM ANOVA analyses for both the involved and uninvolved limbs. No significant differences were comparable across the subjects and only one transition (LGRA) had more than one subject exhibit myoelectric differences within the same transition. Involved limb activation differences were identified only in the quadriceps musculature; predominantly in the VL. The differences occurred in either early stance or early swing. Uninvolved limb differences were identified in shank, quadriceps, and gluteal musculature. These differences were observed at initial heel-strike or from toe-off through swing phase.

Table 6.1
Significant Subject-Specific SPM ANOVA Results for the Involved and Uninvolved Limbs. No subject-specific differences were observed for the LGRD transition. Δ identifies an increase or decrease in myoelectric activation.

	Involved				Uninvolved			
	Subject	**Muscle**	**Gait Cycle**	**Δ**	**Subject**	**Muscle**	**Gait Cycle**	**Δ**
LGRA	3	VL	4-6%	↑	3	MG	4%	↓
	5	RF	23%	↑	-	-	-	-
LGRD	-	-	-	-	-	-	-	-
LGSD	-	-	-	-	4	MG	80-83%	↓
	4	VL	71-74%	↓	4	RF	87-92%	↑
	-	-	-	-	4	VL	50-52%	↓
RALG	-	-	-	-	4	Gmed	0%	↑
RDLG	1	VL	7%	↑	-	-	-	-
SALG	-	-	-	-	2	RF	62%	↑
	-	-	-	-	2	Gmax	57-59%	↑

Common recruitment patterns of the involved limb were observed across transitions. Figure 6.2 shows three representative patterns that were identified in individual subjects.

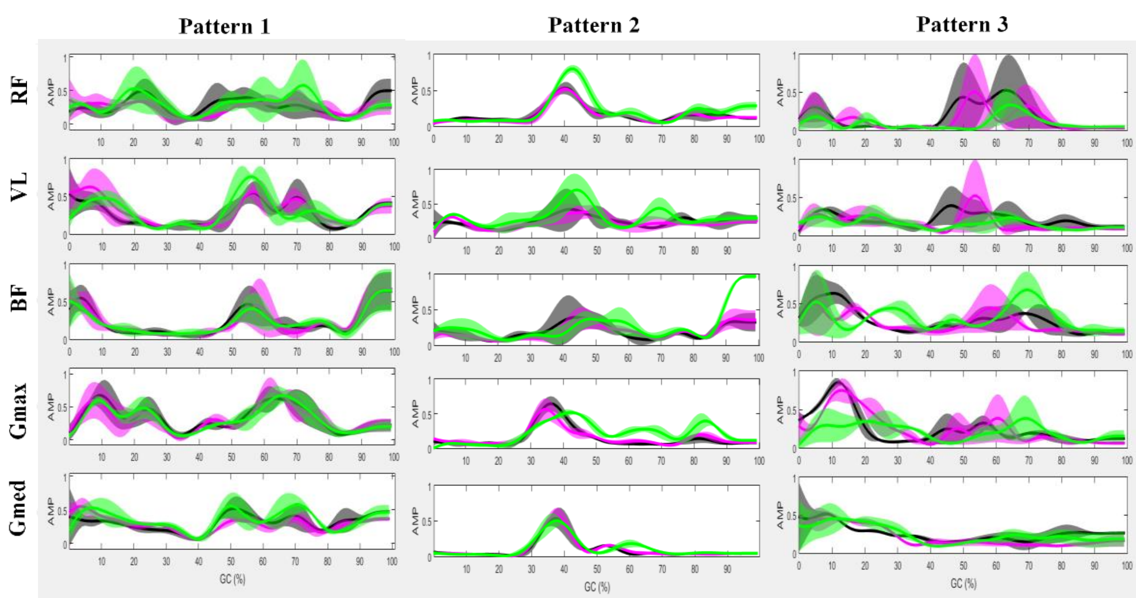


Figure 6.2. Three Common Recruitment Patterns Observed Across the Transitions in the Involved Limb. Myoelectric activation patterns are from three different subjects during the same transition. Patterns 1 and 2 were represented by two subjects each, while pattern 3 was represented by one subject.

Pattern 1 was found to have increased activation during early stance, toe-off, and late swing. Pattern 2 was found to have a primary activation peak during mid-stance with subtle activation through swing. Pattern 3 was found to have increased activation in early stance that diminished through the 40% gait cycle point. Following a brief low activation period, there was an activation peak near toe-off.

6.4 Discussion

Development of a robust EMG controlled powered prosthetic may enable TF amputees to ambulate with minimal consideration for debilitating secondary musculoskeletal pathologies. The purpose of this study was to determine whether lower-extremity EMG in TF amputees could provide sufficient information to classify six different LG, ramp, and stair transitions. It was hypothesized that all transitions would elicit myoelectric differences in at least one of the muscles being studied. The results of the current study do not support this hypothesis.

6.4.1 Predicting Transitions

The lack of identifiable differences in the aggregated data suggest that the general classification of locomotor transitions may not be feasible. The amount of inter-subject variability in TF amputees diminishes the likelihood that a general classification algorithm for locomotor transitions in TF powered prosthetics can be solely governed by lower-extremity EMG. Even with the inclusion of uninvolved limb EMG, only one significant difference was identified. An individualized classification approach may be necessary to develop robust algorithms for TF amputees.

Huang & Ferris (2012) previously concluded that inter-subject EMG variability is high due to individualized recruitment patterns. When EMG was assessed per subject, activation differences began to emerge but were still sparse and varied across muscles and phases in the gait cycle. The lack of repeated significant differences may suggest individualized compensatory mechanisms for encountering a transition. Wentink et al. (2013) showed that TF amputee activation patterns were varied between subjects during LG locomotion. The inclusion of an upcoming transition to differing locomotor terrains may actually enhance the myoelectric differences between subjects.

6.4.2 Myoelectric Activation Patterns

Variable recruitment patterns were found in the involved limb across the five subjects with three distinct patterns emerging in all transitions (Figure 6.2). The patterns primarily seem to lock into common gait events such as heel-strike and toe-off, which corroborate findings by Huang & Ferris (2012) in trans-tibial amputees. It is interesting to note that the within subject patterns seem to occur in not only the transected (thigh) musculature but also in the gluteal musculature. This systemic change to recruitment pattern may be the myoelectric adaptation underlying the kinematic differences observed in previous studies (Bae et al., 2007; Jaegers et al., 1995; Kaufman et al., 2007). For example, TF amputees are known to utilize asymmetric gait to augment the pressure on the residual limb during a hastened stance phase (Bae et al., 2007; Kaufman et al., 2007). Additionally, amputee center of mass trajectories will typically track directly over the prosthesis to improve balance by reducing mediolateral moments on the residual limb (Jaegers et al., 1995). The differing myoelectric patterns identified in the current study may allude to various coping mechanisms during gait retraining.

Three primary myoelectric activation patterns stand out from the involved limb musculature. Pattern 1 exhibits three points of increased myoelectric activation during heel-strike/early stance, toe-off, and late swing. Activation during these phases of the gait cycle may represent an effort to stabilize the prosthesis by bulking the residual musculature within the socket. Similar firing patterns in the gluteal musculature may be secondary to this activation sequence. During heel-strike/early stance, the prosthesis must absorb the energy resulting from initial impact. At toe-off and late swing, it could be presumed that limb bulking would ensure that the prosthesis is secure and will not experience unexpected shifts during limb swing. Alternatively, this pattern could represent pre-activation for the ensuing heel-strike.

Pattern 2 exhibits a single point of activation at approximately 40% of the gait cycle. Activation during this portion of stance is probably too early to be limb bulking for early swing stability as described in pattern 1. The primary purpose of this strategy may be to post the subject's center of mass over the prosthetic side and increase residual limb/socket stability during mid-stance to reduce challenges to balance.

Pattern 3 may be a hybrid of patterns 1 and 2. Increased heel-strike activation slowly diminishes through mid-stance with a second burst of activation around toe-off. Similar to the activation strategies in patterns 1 and 2, pattern 3 may result in greater stability for within the socket during phases of the gait cycle where there is an increased risk of falling.

6.5 Limitations and Future Work

The aim of this study was to determine if a classification algorithm for terrain transitions is feasible using lower-extremity EMG. High variability and varying recruitment patterns in the five subjects yielded inconclusive results. The inclusion of kinematic data would help to validate some of the functional purposes which may be served by the diverse recruitment strategies observed. Furthermore, additional subjects would help to ascertain whether differing recruitment patterns exist beyond the three observed in this study.

Future work should aim to determine whether specific transition differences can be ascertained when partitioning subjects by myoelectric recruitment strategies. It may be possible to design pattern dependent classification algorithms if recruitment strategies are consistent across the broader TF amputee population. To identify the extent of patterns exhibited in the TF amputee population, a principal component analysis may be beneficial, similar to that used by Ivanenko et al. (2004) to identify primary factors able-bodied individuals use while walking and transitioning between differing speeds.

6.6 Conclusion

When assessing the group of five subjects as a whole, only one significant myoelectric activation difference was observed in either the involved or uninvolved limb. When assessing each subject individually, additional significant muscle activation differences were found in locomotion transition conditions. Additionally, individual recruitment patterns emerged which may suggest the need for individualized classification algorithms. However, lower-extremity EMG from TF amputees does not seem to be valuable as a single source of information to govern a classification algorithm for terrain transitions.

CHAPTER VII

SUMMARY OF FINDINGS AND CONCLUSIONS

7.1 Major Findings

The current project aimed to determine if myoelectric differences occurred in able-bodied (Chapter III), TT amputee (Chapters IV and V), and TF amputee (Chapter VI) populations while approaching locomotor state transitions between LG, ramps, and stairs. In all studies, ramp transitions did not elicit activation differences in any of the preceding gait cycles. This suggests that any pre-transition kinematic accommodations programmed into a powered prosthetic controller to facilitate ramp transitions must be designed without the use of lower-extremity pre-transition electromyography.

Able-bodied stair transitions seemed to be primarily accomplished through altered recruitment of the shank musculature. Differences in TA and MG activation were observed in all four stair transitions. Additional differences were identified in more proximal musculature, especially the RF. The combination of the TA, MG, and RF account for movement at the three primary joints in the lower-extremity.

Stair transitions in TT amputees were analyzed for myoelectric activation differences in both the involved and uninvolved limbs. The involved limb produced activation differences in and around the time of toe-off. Residual limb musculature (i.e. TA and MG) produced inconsistent findings. Uninvolved limb musculature produced activation differences in all seven muscles with VL recruitment differences in all four stair transitions. Future designs of classification algorithms for TT amputees should consider utilizing the involved limb activation differences around toe-off, uninvolved limb thigh musculature activation differences, and uninvolved limb activation differences during swing phase. The lack of residual limb differences may suggest low utility in enhancing classification schemes.

Locomotor state transitions in TF amputees were similarly analyzed for myoelectric activation differences in both the involved and uninvolved limbs. From the SPM ANOVA analysis of sample data, only VL in one transition was identified to change pre-transition. When subjects were analyzed individually, three distinct myoelectric recruitment patterns in the involved limb musculature were observed. Interestingly, the patterns were observed in all muscles of the involved limb and not just those in the residual limb (i.e. RF, VL, and BF). These patterns may suggest independent compensation strategies learned during gait re-training. Validation of these recruitment patterns and subsequent studies assessing grouping TF by the identified patterns may present transition-by-transition activation differences. At present, it does not seem feasible to classify transitions solely based on lower-extremity EMG in TF amputees.

7.2 Limitations

Implementation of a classification algorithm for lower-extremity powered prosthetics implies the use of powered prosthetics by amputees. However, the current study assessed EMG from amputees who were using passive prosthetics. This represents the primary limitation of this project. The use of passive prosthetics by lower-extremity amputees for use in EMG-based algorithms is well documented (Huang et al., 2011, 2009; Huang and Ferris, 2012; Johnson et al., 2014; Wentink et al., 2013). Studies of amputees who wear passive prosthetics are common because of the lack of economically feasible powered prosthetics in the open market. Furthermore, lower-extremity powered prostheses remain difficult to control, which offsets the proposed benefits. Thus, passive prosthetics are currently used to understand how amputee gait differs from able-bodied gait. However, with the return of internal knee and/or ankle moments, powered prosthetics may completely alter lower-extremity muscular recruitment.

The use of a single dimension for both the ramp and stair terrains could be considered another limiting factor of this project. The ramp angle was measured to be at a 5° incline/decline, while the stair terrain was measured to be 16.5cm tall x 30.5cm deep. A previous study concluded that recruitment patterns of lower-extremity EMG in able-bodied subjects changed proportionally with ramp slope (Sheehan and Gottschall, 2012). It was observed that a medium slope of ±9° or greater significantly altered EMG activation. At that value, the slope would exceed the standard set by the Americans with Disabilities Act (ADA). The stair dimensions are in full compliance with Occupational Safety and Health Administration (OSHA) bylaws and Building Officials and Code Administrators (BOCA) guidelines. These entities, partially or in full, govern the design of public access buildings and structures. Though these limits do not account for all situations, they do account for a majority of the situations one may encounter in daily life.

A set of seven muscles, when available, were assessed through the duration of this project. This set of musculature was chosen to represent most major movements in the lower-extremity as they relate to LG, stair, and ramp locomotion. However, additional muscles could be analyzed that may highlight different and/or better transition related differences. Through musculoskeletal modeling, Steele et al. (2013) showed how muscle selection impacts the variability of predicting muscle synergies. Similar to how a classification algorithm could operate, careful consideration must occur when deciding which and how many muscles will be used in an algorithm.

7.3 Suggestions for Future Work

This project begins to identify how lower-extremity muscular activation changes in response to an upcoming locomotor transition in the able-bodied, TT amputee, and TF amputee populations. Based on the findings of this project, additional areas of research have presented themselves.

Data collected from the TF amputee sample did not result in findings that were similar to either the able-bodied or TT amputee samples. The findings from Chapter VI suggest that there may be different recruitment patterns being employed by TF amputees. A previous study looking at EMG activation patterns while transitioning between different speeds identified five different locomotor recruitment patterns using Principal Component Analysis (Ivanenko et al., 2004). This technique could be utilized with data from a TF amputee sample to determine if distinct involved limb musculature recruitment patterns exist. If so, TF amputees can be grouped by recruitment pattern then assessed as locomotor state transitions are approached.

This project was conducted with the goal of providing insight for improving EMG-based classification algorithms. Findings from able-bodied and TT amputee data suggest that lower-extremity EMG could be used to identify upcoming transitions. Algorithms developers should use the activation differences identified in this project to determine the predictive value in classifying upcoming transitions. Additional research is needed to determine if using multiple sources of information would further strengthen the validity and reliability of the classification schemes. Inertial measurement units may provide valuable information as a fine-tuning mechanism for the classification efforts.

As powered prosthetic use becomes more common in the lower-extremity amputee population, further research should be conducted to determine if myoelectric activation patterns are similar to able-bodied activation patterns, passive prosthetic activation patterns, or present a different activation pattern altogether. Considering the scenario of amputees switching from passive to powered prosthetic use, research should also aim to determine the plasticity in the myoelectric activation patterns as symmetrical gait returns.

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