Indicators of Anticipated Walking Surface Transitions for Powered Prosthetic Control

by

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ABSTRACT

Human locomotion is an essential function that enables individuals to lead healthy, independent lives. One important feature of natural walking is the capacity to transition across varying surfaces, enabling an individual to traverse complex terrains while maintaining balance. There has been extensive work regarding improving prostheses' performance in changing walking conditions, but there is still a need to address the transition from rigid to compliant or dynamic surfaces, such as the transition from pavement to long grass or soft sand. This research aims to investigate the mechanisms involved such transitions and identify potential indicators of the anticipated change that can be applied to the control of a powered ankle prosthetic to reduce falls and improve stability in lower-limb amputees in a wider range of walking environments. A series of human subject experiments were conducted using the Variable Stiffness Treadmill (VST) to control walking surface compliance while gait kinematics and muscular activation data were collected from three healthy, nondisabled subjects. Specifically, the kinematics and electromyography (EMG) profiles of the gait cycles immediately preceding and following an expected change in surface compliance were compared to that of normal, rigid surface walking. While the results do not indicate statistical differences in the EMG profiles between the two modes of walking, the muscle activation appears to be qualitatively different from inspection of the data. Additionally, there were promising statistically significant changes in joint angles, especially in observed increases in hip flexion during the swing phases both before and during an expected change in surface. Decreases in ankle flexion immediately before heel strike on the perturbed leg were also observed to occur simultaneously with decreases in tibialis anterior (TA) muscle activation, which encourages additional research investigating potential changes in EMG profiles. Ultimately, more work should be done to make strong conclusions about potential

indicators of walking surface transitions, but this research demonstrates the potential of EMG and kinematic data to be used in the control of a powered ankle prosthetic.

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Chapter 1

INTRODUCTION

Human walking is an important but intricate function that coordinates balance and enables mobility, which in turn allows for other essential tasks and a promotes an independent lifestyle. However, many individuals suffer from impaired gait. Nearly 2 million Americans are estimated to be living with limb loss, and the majority of these cases are lower-limb amputations. Conventional prosthetics can restore walking capabilities but have yet to replicate natural walking in more complicated walking conditions. This chapter will introduce the motivation of this research, expand on its need, and clarify its specific objectives.

1.1 Motivation

Limb loss is estimated to affect approximately two million people in the United States alone and increase to affect over 3.6 million Americans by the year 2050 (Ziegler-Graham et al. 2008, 424). The leading cause for amputations in these cases is dyvascular disease, which is increasingly becoming a larger contributor to limb loss in the United States. For instance, while dyvascular disease accounts for 54% of current instances of limb loss, it is also attributed to 80% of new amputations (Pasquina et al. 2014, 273).

Rehabilitation is a serious concern for those living with limb loss. Mobility is important to living a healthy lifestyle, but regaining walking ability and balance can be a challenge after lower-limb amputations, even with the aid of prosthetics. Lower-limb amputees report low balance confidence, especially in those whose amputations were related to vascular disease (Miller, Speechley, and Deathe 2002, 856). In a study conducted by Miller et al. (2001, 1238), over half of individuals with lower-extremity amputations reported falling in the past 12 months, and almost half reported a fear of falling. The consequences of falling is amplified since many individuals with lower limb loss, and especially in amputations caused by vascular disease, are older in age and thus more susceptible to serious injury in a fall. Simply the fear of falling can affect quality of life. According to Miller, Speechley, and Deathe (2002, 856), balance confidence is strongly correlated to social activity, which further emphasizes the need to reduce or eliminate the fear of falling.

1.2 Objective and Scope

The use for a powered ankle prosthetic capable of navigating all types of terrains, including surfaces of changing compliance, is clearly illustrated and poses an interesting challenge. However, to prevent falls, it is desirable for the prosthetic to actively adjust prior to encountering a complex surface. The objective of this research is to evaluate gait mechanisms during surface transitions and to identify potential indicators of an anticipated walking surface change, which could then be applied to the control of the powered prosthetic. The scope of this research will be limited to investigating the muscle activation associated with ankle flexion, which will be done using the VST, a novel split-belt tool that is able to change the compliance of its left walking surface, i.e. perturb the surface, in real time. The hypothesis is that the EMG profile of the perturbed leg significantly changes during the step immediately before an expected perturbation. If proven true, this would suggest that the residual limb muscle activity of an transtibial amputee could be used as an input to a powered prosthetic.

Chapter 2

LITERATURE REVIEW

Human gait is made possible by the culmination of several complex processes which, for example, use gathered sensory and proprioceptive feedback, combined with streamlined muscle activation and inter-limb coordination, all while maintaining balance and posture. One of the most useful outcome of these intricacies is the ability to adapt to and navigate through a wide range of terrains. Ultimately, this research aims to restore this capability to individuals with lower limb loss by contributing to the design of a powered prosthetic adaptable to varying walking conditions. This chapter will evaluate studies that expand the understanding of human adaption to dynamic walking surfaces as well as review previous works on powered ankle control and performance.

2.1 Gait Adaptions on Complex Surfaces

From lush green grass to soft white sands to and hard, jagged rocks, humans are able to traverse a variety of walking surfaces that impressive feats, which even include summiting Mt. Everest and walking on the moon. However, this ability is also essential for simple tasks, facilitating many day-to-day activities such as crossing a lawn or climbing stairs. Typically, conservative measures such as increasing toe clearance are taken to improve stability on complex surfaces and reduce the likelihood of falls. It is useful to understand these mechanisms before attempting to reproduce this natural behavior in any prosthetic.

Common trip hazards encountered in walking include changes in height, levelness, or slope. Extensive research has been conducted to understand how gait mechanisms change to adapt to this complication and still maintain balance and cadence. It is reasonable to expect changes to gait to accommodate new obstacles, which has been confirmed in several studies. Estermann et al. demonstrated that there is a shift in the synchronization of muscle activation while walking on a cross-sloped surface compared to level walking, demonstrating the ability of small alterations to walking surface to significantly alter gait patterns (2016, 61). This is also seen in uneven terrain, as in the study conducted by Wade et al. investigating gait dynamics on irregular sloped surfaces using railroad ballast to create an uneven surface (2014, 1658). The results determined that walking on the ballast required significantly larger joint moments to be generated. Moreover, the results exemplified the difficulties of walking on such terrain, which is a regularly encountered occupational environment for many people.

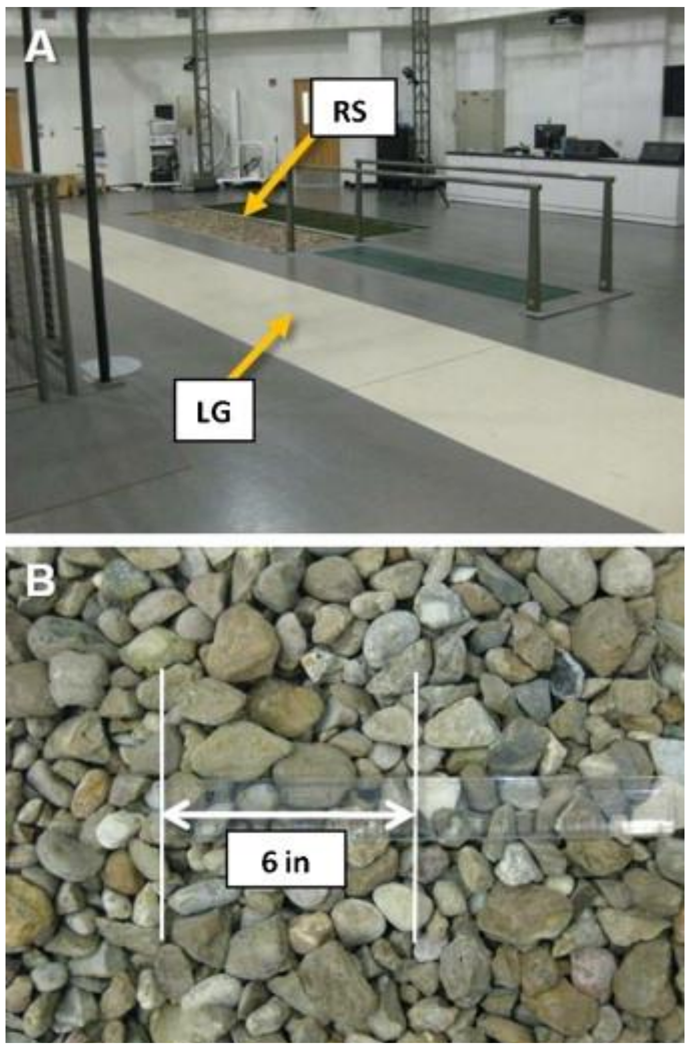

Figure 1: Experimental Setup Used by Gates et al. (2012a, 37).

Of course, individuals suffering from limb loss are no exception to the effects of uneven ground. Gates et al. (2012a, 33-39) noted that subjects with transtibial amputation, using energy-storing prosthetic feet, took notable, conservative measures while walking on a destabilizing rock surface. The experimental setup used in this study is presented in Figure 1. The subjects took shorter strides with a wider stance and increased toe clearance by increasing hip and knee flexion in the intact leg and hip flexion in the prosthetic leg. However, the adaptions were asymmetrical due to the amputation compared to the gait changes previously observed in individuals with no amputations (Gates et al. 2012b, 36-42). This comparison highlights the disadvantages of using conventional protheses in complex surfaces and supports the need to improve the state of powered prosthetics to better mimic natural gait responses.

However, while these works are useful for understanding the gait mechanisms involved in walking over some common obstacles, they were limited to hard, rigid surfaces, which is only a subset of natural environments an individual might encounter. Other types of surfaces that are increasingly gaining attention are compliant and dynamic surfaces. These are common in outdoor environments, where the walking surface displaces while being walked on. Real-world examples include dirt, grass, and sand. These surfaces are also found to significantly alter gait mechanics, as evidenced in prior studies. MacLellan and Patla investigated the leg kinematics and the muscle activation of healthy subjects walking on a compliant mat surface (2006, 521-30). A depiction of the experimental setup is given in Figure 2. The study yielded similar results to the previously mentioned works; subjects altered their step width and length and increased toe clearance on the compliant surface. In addition, increased activation of the gastrocnemius and soleus muscles were observed during push-off, which contributed to the difference in step length. Typically, the tibialis anterior (TA), which is responsible for ankle dorsiflexion, would be expected to increase as toe clearance increases, but these results identifies these muscles as being significant to the gait mechanisms used in walking over compliant surfaces and thus may also be important to the transition between surfaces as well.

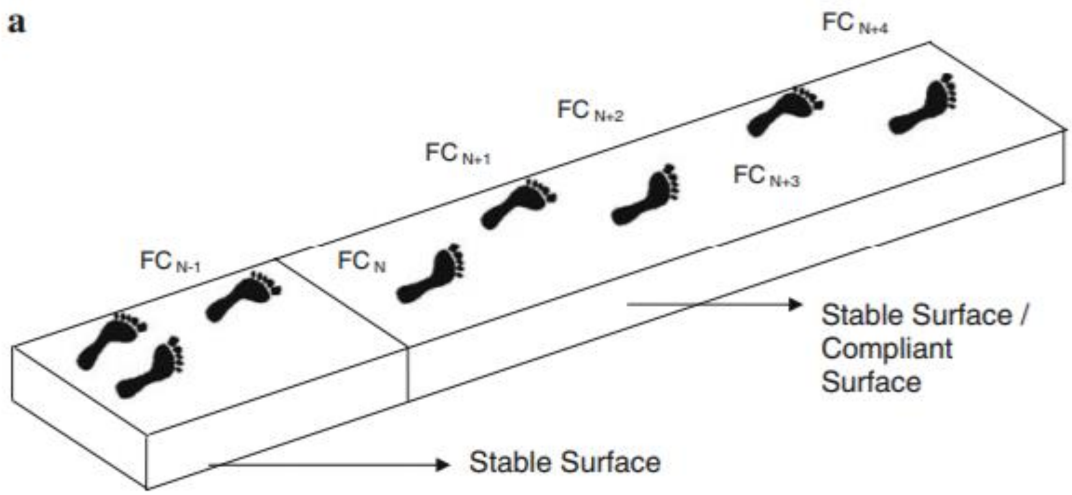

Figure 2. Experimental Setup by MacLellan and Patla (2006, 523).

Many of the studies discussed thus far utilize controlled environments in their experiments, which is useful for simulating some aspects of real-world walking surfaces and understanding gait adaptions to the specific parameters investigated in those studies. Beyond this, it is also appropriate to examine the mechanisms of gait on actual, real-world surfaces instead of only their simulated counterparts. In a study conducted by Paysant et al., the gait parameters for both nondisabled persons and transtibial amputees, using energy-storing prostheses, were investigated for walking across three kinds of common, real-world terrains: asphalt, mown lawn, and high grass (2006, 153-60). The results did not find significant changes between asphalt and mown lawn for either subject group, but there were some differences in gait between groups on asphalt and mown lawn, primarily in walking speed and oxygen cost. This indicates that the difference in surface characteristics between asphalt and mown lawn is not great enough to evoke a gait adaption but that generally, for even simple surfaces such as asphalt, there is a difference in the walking ability between nondisabled and amputees. Specifically, the disadvantages of the amputee group are a slower, more conservative walking speed and a larger metabolic cost. There were also significant differences in all metabolic and temporal parameters between asphalt and high grass and between nondisabled and amputee groups. That is, not only did walking on high grass lead to gait changes in both groups compared to walking on asphalt, but the changes were more drastic in the amputee group compared to the nondisabled group, again highlighting the disadvantages of amputees using conventional prosthetics. Generally, these findings support the need for improvements in powered prosthetics to help individuals with limb loss traverse different surfaces more naturally.

2.2 Classifications of Locomotion Modes

Finally, it is also important to identify mechanisms involved in transitioning between types of terrain during walking. This presents an interesting problem, and accordingly, many researchers have attempted to classify different locomotion modes and transitions. Miller, Beazer, and Hahn developed a myoelectric walking mode classifier for transtibial amputees that attempted to discriminate between level ground walking, stair ascent and decent, and ramp ascent and descent using surface EMG sensors to measure muscle activation in muscles pertaining to knee and ankle flexion in both amputee and nondisabled groups (2013, 2746-2767). The placement of sensors is demonstrated in Figure 3. The study achieved at least 96.4% accuracy in differentiating between the five major locomotion modes studied, with the most accurate being the classification of stair ascent and descent. This demonstrates the potential to identify different terrains using EMG data; however, one major drawback of the study is that it focused only on steady-state walking on the various surfaces. Ideally, this thesis will identify gait features that indicate a change in walking surface prior to the actual transition because it is desirable for a powered prosthetic to begin adapting before encountering a new terrain, which will improve stability, lower fall risk, and reduce any compensation needed post-contact.

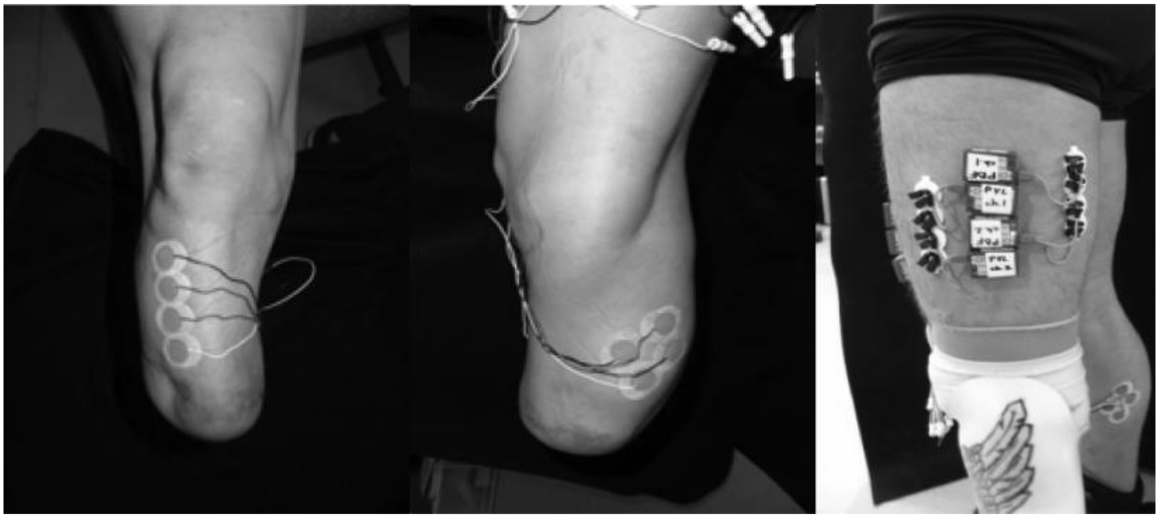

Figure 3. Example Sensor Placement by Miller, Beazer, and Hahn (2013, 2746).

In comparison, Huang et al. attempted to classify the transitions from level ground walking to stair and ramp ascents and obstacle clearance using a combination of EMG and mechanical information from transfemoral amputee subjects (2011, 2868-2869). Ultimately, they yielded 100% accuracy in classifying all transitions across every single subject using a fusion-based supports vectors machine (SVM) method. Still, a subsequent study conducted by Joshi and Hahn (2013, 1275) aimed to address drawbacks in the study performed by Huang et al. According to Joshi and Hahn, the previous study had gaps in that it failed to study the transition to descent modes, such as descending on stairs or ramps and was not able to collect mechanical data during the swing phase due to use of a load cell (2013, 1275). This study aimed to address those drawbacks by using accelerometers and switches to obtain mechanical information and gait events as well as include descent modes in the investigation. An example of the sensor placement used and a sample transition gait cycle is represented in Figure 4. However, they yielded less accurate classifications, only achieving up to 95.2% accuracy in the best case: identifying

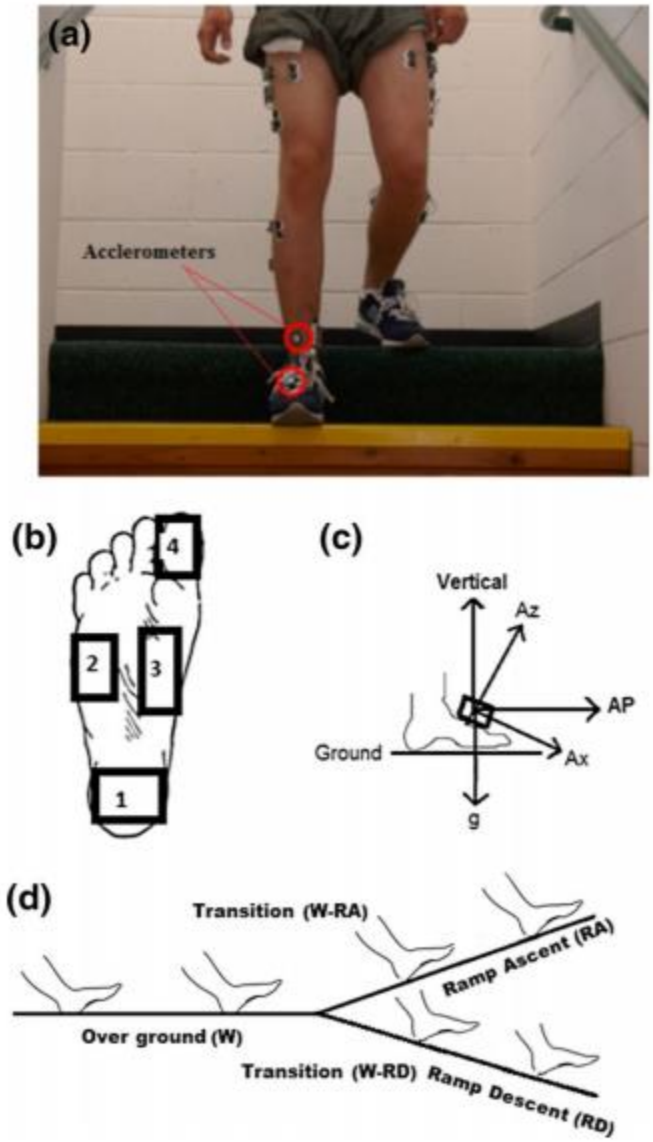

Figure 4. Example Sensor Placement by Joshi and Hahn (2013, 1276).

direction type transitions, i.e. differentiating the transitions from level walking to ascent modes, using either stair or ramp, and from level walking to descent modes. One challenge outlined in their findings was the uncertainty in separating locomotion modes. It was difficult to discern when a transition should occur, and it was noted that the distance between toe off and the physical surface change may play a role in when transitional behavior occurs. For example, Joshi and Hahn hypothesize that if the toe off occurs farther away from the actual surface transition, then a large portion of the gait cycle will follow the trends of the previous locomotion mode compared to instances when the toe off occurs relatively close to the physical transition location. This emphasizes the need to regulate the occurrence of the walking surface change with respect to the gait cycle and supports the VST as an effective investigational tool since perturbations can be programmed to occur during specific portions of the gait cycle.

Overall, the studies on mode classification give evidence that it is possible to identify upcoming transitions in walking surfaces. Yet, there is an obvious gap in research in this area. While the studies mentioned investigate changes in level ground walking to other rigid surfaces such as stairs and ramps, they all fail to address changes in surface compliance. There may be significant difference in locomotion between these modes since the primary difference between level ground walking and ascending or descending stairs and ramps are the change in surface height and grade. In contrast, a change in compliance is dynamic but does not necessarily begin with a change in slope or level, eventually resulting in vertical deflection and height compensation after contact with the surface. Furthermore, there is more likely to be energy lost to an encounter with a compliant surface, thus requiring more leg work in compensation. Thus, there is strong potential for the gait mechanisms required during transitions with compliant surfaces to be different than the modes discussed in these works.

Chapter 3

METHODOLOGY

To test the hypotheses presented in the objective, it is desired to design an experiment that achieves the following:

- Allows for the collection of leg kinematics and muscle activation data.
- Exposes the subject to surfaces of varying compliance during walking.
- Produces predictable, anticipated transitions between walking surface compliance.
- Standardizes surface compliance(s) and transition timing for all subjects.

This chapter will detail the methods used to satisfy the above requirements during experimentation.

3.1 Experimental Setup

Considering the conditions mentioned above, the Variable Stiffness Treadmill (VST) developed by Skidmore, Barkan, and Artemiadis is chosen as the primary tool of investigation and used to employ patterned, anticipated stiffness perturbations to examine the gait mechanisms that occur when a change in walking surface is expected (2014, 1717-24). The VST was developed with the ability to change the surface compliance during walking in real-time using its novel variable stiffness mechanism. In addition, these changes in compliance can be programmed for a walking experiment such that they can be anticipated by the subject. It provides several advantages such as a wide range of controllable and accurate walking surface stiffness and consistent perturbation timing relative to a user's gait cycles. The system is comprised primarily of the variable stiffness mechanism, split-belt treadmill, and motion capture system. There are also a body weight support (BWS) and harness system and handrails for safety. These components will be discussed in further detail in the upcoming sections. Some components and their setup are depicted in Figure 5.

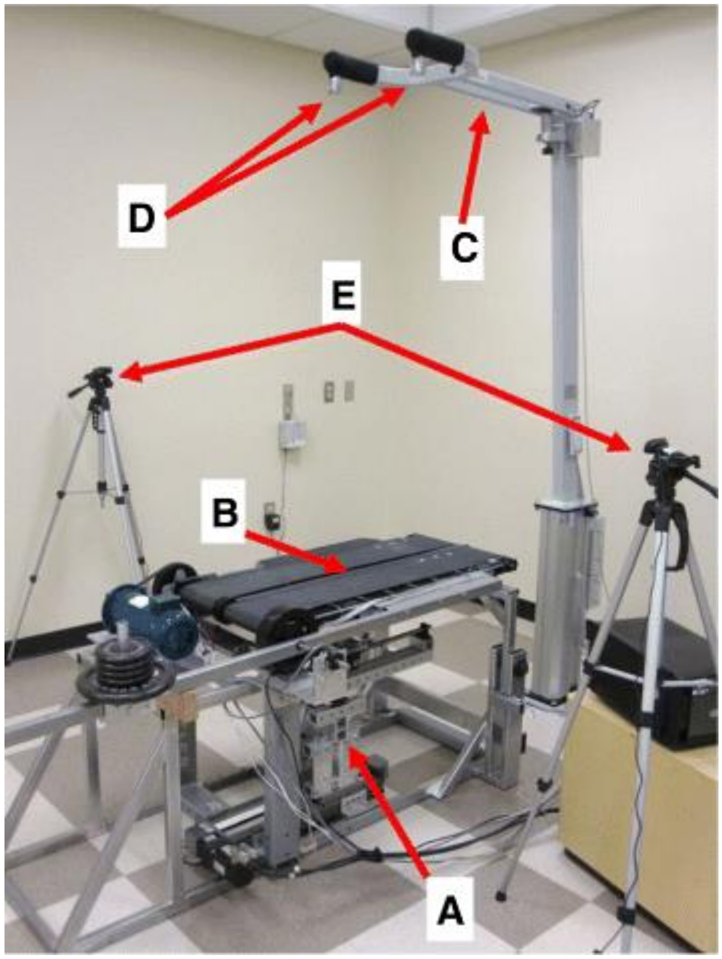

Figure 5: Variable Stiffness Treadmill (VST) Setup. (A) Variable Stiffness Mechanism, (B) Split-belt Treadmill, (C) Harness-Based Body-Weight Support (BWS), (D) BWS Loadcells, (E) Motion Capture System.

3.1.1 Variable Stiffness Mechanism

The VST can vary the effective stiffness of its walking surface using its unique stiffness mechanism. The mechanism consists of a spring-loaded lever that translates on a linear, horizontal track installed underneath the treadmill. When vertical force is applied to the treadmill, as when a user steps onto the belt, the mechanism provides a reciprocal force that is dependent on the stiffness S of the linear spring and its position x on the track, which alters the length of the resultant moment arm. Thus, the effective stiffness of the treadmill is changed by repositioning the stiffness mechanism on the track. This concept is exemplified in Figure 6. The VST system employs a high-capacity linear track (Thomson Linear, Part Number: 2RE16-150537) and a precision drive (Kollmorgen, Part Number: AKD-P00606-NAEC-0000). Further details on the development and analysis of this mechanism is provided by Skidmore, Barkan, and Artemiadis (2014, 1717-24).

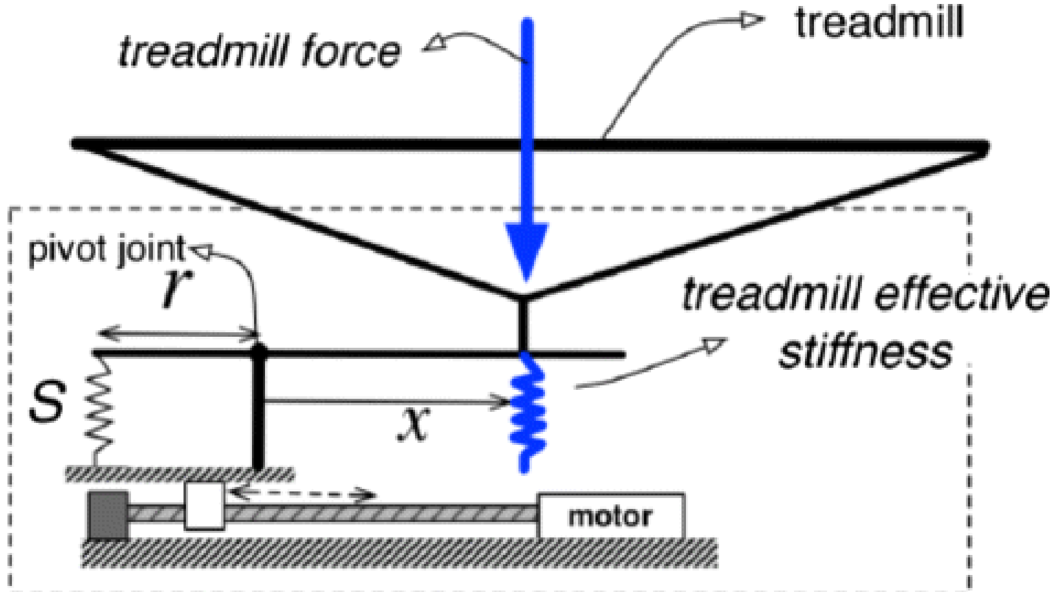

Figure 6: Conceptual Diagram of Variable Stiffness Mechanism.

3.1.2 Split-Belt *Treadmill*

As depicted in Figure 5, the walking surface consists of a split-belt treadmill. The treadmill belts are supported 70 cm above the floor to accommodate the stiffness mechanism and enable each belt to independently deflect. However, since the stiffness mechanism is installed below the left treadmill belt only, the stiffness of the left leg walking surface can be independently varied while the right treadmill belt will always remain rigid. This can be a limitation of the system, but since this research primarily focuses on investigating gait immediately before and during a perturbation, only one leg is required to be perturbed, and this disadvantage is deemed acceptable, especially compared to the many previously described benefits of using the VST.

3.1.3 Motion Capture System

Leg kinematics is recorded by using a system of infrared (IR) LED markers (Super Bright LEDs Inc, model: IR-1WS-850) and IR cameras (Code Laboratories Inc, model: DUO MINI LX). Together, these create a low-cost motion capture system. Six markers are on secured laterally onto each leg--two on the foot, two on the shank, and two on the thigh--such that they are parallel to the subject's sagittal plane. Two IR cameras are positioned on either side of the VST, aimed at the IR markers when an individual is on the treadmill. Prior to beginning experiments, they are calibrated to the length and position of the treadmill to ensure accurate position data is collected. At the start of an experiment, the cameras record the initial position of markers while the subject is standing still. This position data is used to calculate the initial, or “zero,” hip, knee, and ankle joint angles. During the experiment, the cameras track the positions of the markers as the subject is walking, which are then used to recalculate the joint angles relative to the initial position in real time. Figure 7 shows the setup of the VST, IR equipment, and a representative subject wearing the IR markers.

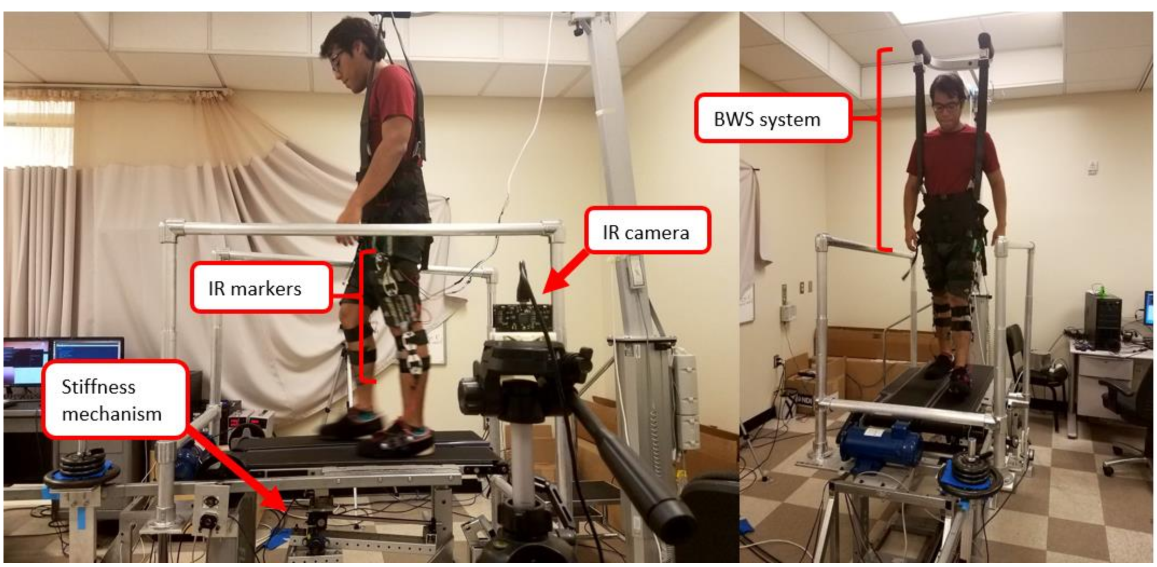

Figure 7. Experimental Setup with Representative Subject

The kinematics data is not only useful for investigating gait mechanisms, but this data is also used to control the VST and time the changes in surface compliance with respect to the subject's gait cycle. The gait cycle is assumed to begin at the left leg heel strike and continues throughout the stride until the next left leg heel strike. During this cycle, the left leg is in contact with the ground for approximately 60% of the gait cycle, from the initial heel strike to toe-off, which is also called the stance phase. During the remaining 40% of the gait cycle, the left leg is in what is considered the swing phase and is moving forward in the air in preparation for the next gait cycle. To best simulate real-world walking environments, the change in compliance should occur during this swing phase so that the change in compliance is experienced by the user at the beginning of the next gait cycle. In practice, normal walking surfaces do not suddenly change compliance under an individual's foot while in the stance phase. Instead, walking surface changes normally occur between steps, which should be replicated in the VST. Since the resulting kinematics data provides foot position, the swing phase is identified by tracking the direction that the foot is moving in. When the foot is determined to be moving forward, i.e. in the swing phase, then the VST's stiffness mechanism changes in order to provide the appropriate stiffness for the next gait cycle.

3.1.4 Wireless Electromyography (EMG) System

In conjunction with the collection of kinematics data, the muscle activity of both legs is also obtained using electromyography (EMG) and recorded using a wireless surface EMG system (Delsys, Trigno Wireless EMG). This study focuses on three muscles on each leg pertaining to ankle flexion: the tibialis anterior (TA), gastrocnemeus (GA), and soleus (SOL). Before the experiment, correct placement of the sensors is confirmed by observing the muscle activation of each muscle while the subject is told to execute each muscle’s associated motion, e.g. dorsiflexion for the TA and plantarflexion for both the GA and SOL. The placement of the electrodes is demonstrated in Figure 8.

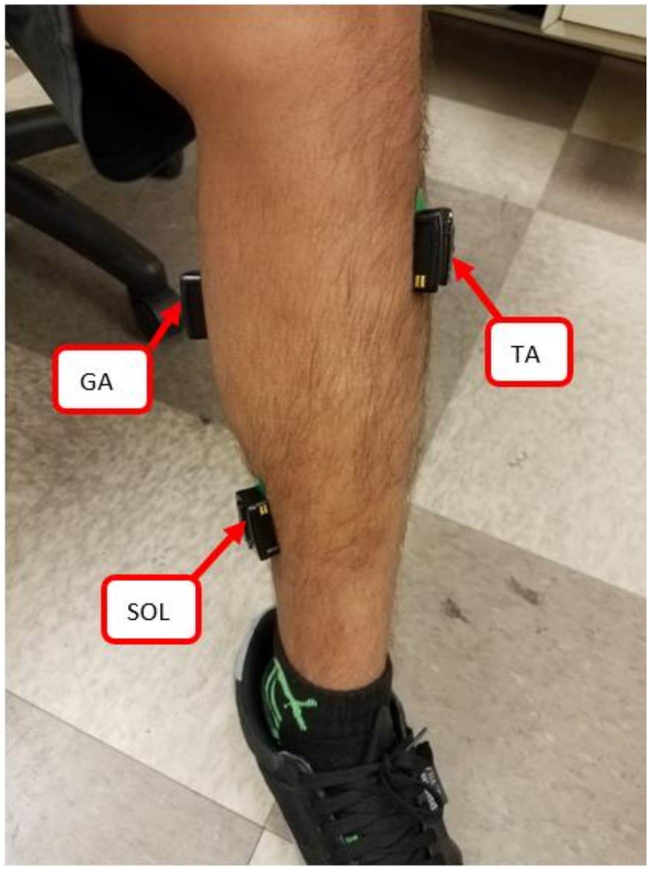

Figure 8: Example EMG Sensor Placement.

3.2 Experimental Protocol

To evaluate potential indicators of expected walking surface changes, the VST is used to alter the compliance of the treadmill during walking. The allocations and timing of the perturbations are pre-determined and patterned, enabling the subject to be notified of upcoming changes in walking surface during the experiment.

The experiment begins with 30 gait cycles at infinite stiffness to help establish a baseline profile for rigid ground walking. Following this introductory period, the treadmill will apply alternating sets of three successive perturbed gait cycles at a relatively low stiffness and three successive rigid gait cycles. During the perturbations, the stiffness mechanism is commanded to an x position of 4.0 cm from its initial position on the track, which results in an effective stiffness range of 25-30 kN/m. The perturbed-rigid cycle pattern alternates for a total of 30 times, after which there are an additional 30 gait cycles at infinite stiffness to conclude the experiment. Table 1 shows the allocation of perturbations throughout the experiment.

Table 1: Allocation of Perturbations During Experiment.

Gait Cycle(s)	Number of Cycles	Walking Mode
1-30	30	Normal
31-33	3	Perturbed
34-36	3	Normal
37-39	3	Perturbed
40-42	3	Normal
⋮	⋮	⋮
205-207	3	Perturbed
208-210	3	Normal
211-240	30	Normal

Because the perturbations are programmed to occur during known gait cycles throughout the experiment, the number of gait cycles until the next perturbation can be easily determined. This is used to verbally notify the subjects of an approaching perturbation. Specifically, subjects were told how many gait cycles, or steps, were left before the next surface change was to occur, counting down to "zero," which indicates that the next immediate step will be perturbed. Moreover, the repetitive pattern of the perturbations employed during the experiment further help the subject anticipate upcoming transitions.

Three healthy, nondisabled subjects (age = 22±2.1 years, weight = 667±20.8 N) with no walking impairments volunteered to participate in this study. All subjects provided their informed consent. Although no BWS was used in this experiment, every subject was required to wear a harness for safety. This experiment is approved by the Arizona State University Institutional Review Board (IRB#: STUDY00001001).

3.3 Data Analysis

As previously mentioned, leg kinematic data, particularly foot position and joint angles, and EMG for both legs were collected in real-time during the experiments.

Kinematic data was sampled at 140 Hz using the motion capture system discussed in 3.1.3. This data was crucial in determining the timing of gait cycles, which is used to separate individual gait cycles and further analyze other data components. The gait cycle is taken to begin at the left leg heel strike, which is considered to occur when the tracked foot position reaches its minimum, i.e. when the foot has fully extended forward following the swing phase. The cycle continues throughout the subject's stride until the next detected heel strike. After determining the instances of left leg heel strike, it is possible to determine the duration of each gait cycle and section the data according to gait cycles of interest.

Muscle activation was obtained using the wireless EMG system detailed in 3.1.4, and the resulting raw EMG signals were recorded at 2000 Hz. The amplitudes of these signals were estimated by finding the root mean square (RMS) envelope of the signals corresponding to each muscle within 250 ms windows using Simpson's 1/3 rule, then normalized to the maximum values of the estimated amplitudes.

The kinematic and EMG data for each gait cycle was resampled at each 0.1% of the gait cycle. Since it is desired to investigate the gait cycles immediately before and during an anticipated walking surface transition, the data was sectioned into windows of two gait cycles and further separated by windows containing a surface stiffness change and windows only containing rigid walking. The validity of the separation methods is confirmed in Figure 9, which demonstrates the average difference in treadmill deflection between windows containing a surface transition and normal walking. In this figure, as well as in the following figures in the next chapter, the first and second gait cycles correspond to 0-100% and 100%-200% gait cycle on the x-axis, respectively. Moreover, for windows representing a walking surface transition, the first gait cycle is the stride immediately before an expected change in surface stiffness while the actual perturbation occurs in the second gait cycle. In Figure 9, it is apparent that there is relatively small difference in treadmill surface deflection during the first gait cycle, which should be very close to infinite stiffness in both types of walking. Then, during the second gait cycle, it is obvious that the treadmill significantly deflects during a walking surface transition, which aligns with expectations since a perturbation occurs during this cycle for anticipated surface changes. This validates the methods used to segment the data. A black vertical dotted line is also present to indicate where the treadmill deflection becomes significantly large during a perturbation compared to normal walking. This occurs at approximately 110% gait cycle for all subjects.

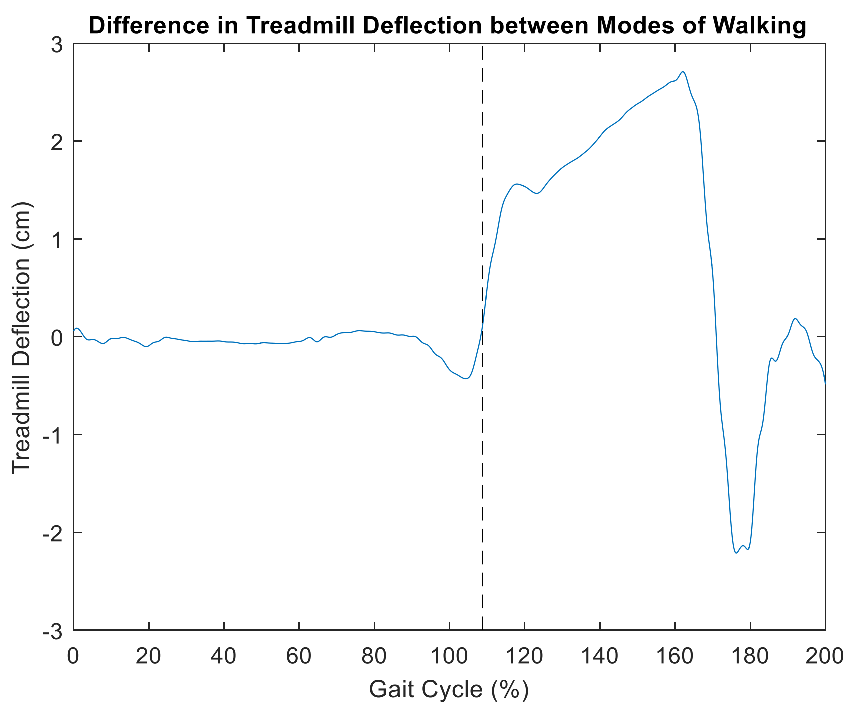

Figure 9: Plotted Difference in Treadmill Deflection between Modes of Walking.

Both kinematic and EMG data were averaged amongst all the windows for each type of walking (stiffness transition and rigid walking). In addition, two-tailed, two-sample unpaired t-tests were performed to compare the data between these two modes and quantify statistical significance. This test was chosen since it is desired to compare two independent distributions with different sample sizes: gait cycles that include and do not include a walking surface transition. Statistical significance was concluded if the test indicated significance for at least 4% of the gait cycle. The tests were performed at a 95% confidence level, the results of which are shown and discussed in the following chapter.

Chapter 4

RESULTS AND DISCUSSION

This chapter presents and discusses the results of the experiment described in Chapter 3. The following plots compare the joint angles between cycles where an anticipated perturbation occurs and cycles with only normal, rigid ground walking for both left and right legs (Figures 10-11), as well a similar comparison for muscle activation in both legs (Figures 12-13). These figures give the results for representative subjects.

4.1 Preliminary Observations

4.1.1 Effects on Perturbed Leg Muscle Activation and Kinematics

Generally, the focus of this research is on the gait mechanisms of the perturbed leg, or in this case, the left leg. It was hypothesized at the beginning of this study that the EMG profile of the perturbed leg will be significantly different immediately before and at the beginning of an anticipated change in walking surface stiffness. This would have exciting implications for powered prosthetic control since this suggests that the residual limb muscle activation of a lower-limb amputee could potentially be used to adapt a prosthetic to compliant surfaces. In addition, this would be ideal from a design perspective since the necessary sensors would be localized to the residual limb, close to the physical prosthetic and minimally intrusive to the end-user.

A comparison of EMG profiles from a representative subject is given in Figure 10. From initial inspection, there are apparent changes in EMG profiles during the perturbation, which is consistent with previous studies of walking on compliant surfaces (MacLellan and Patla 2006, 521). For example, there are pronounced increases in the GA and SOL activation, which is expected as the subject increases push off to compensate for the now-compliant surface. One interesting observation is that TA activation appears to decrease during the swing phase immediately before the perturbation occurs and into the beginning of the next heel strike, in the range of approximately 70-120% gait cycle. This trend was also present among other subjects and may be indicative of the subjects preparing to adapt to the perturbation.

While statistical analysis was performed to search for quantifiably significant differences in EMG profiles, test results indicated that there were no statistically significant differences while assuming unequal variances at any part of the gait cycle. This was consistent for all subjects. The lack of statistical significance could stem from the relatively large variance in EMG data, depicted by the shaded area in the figure.

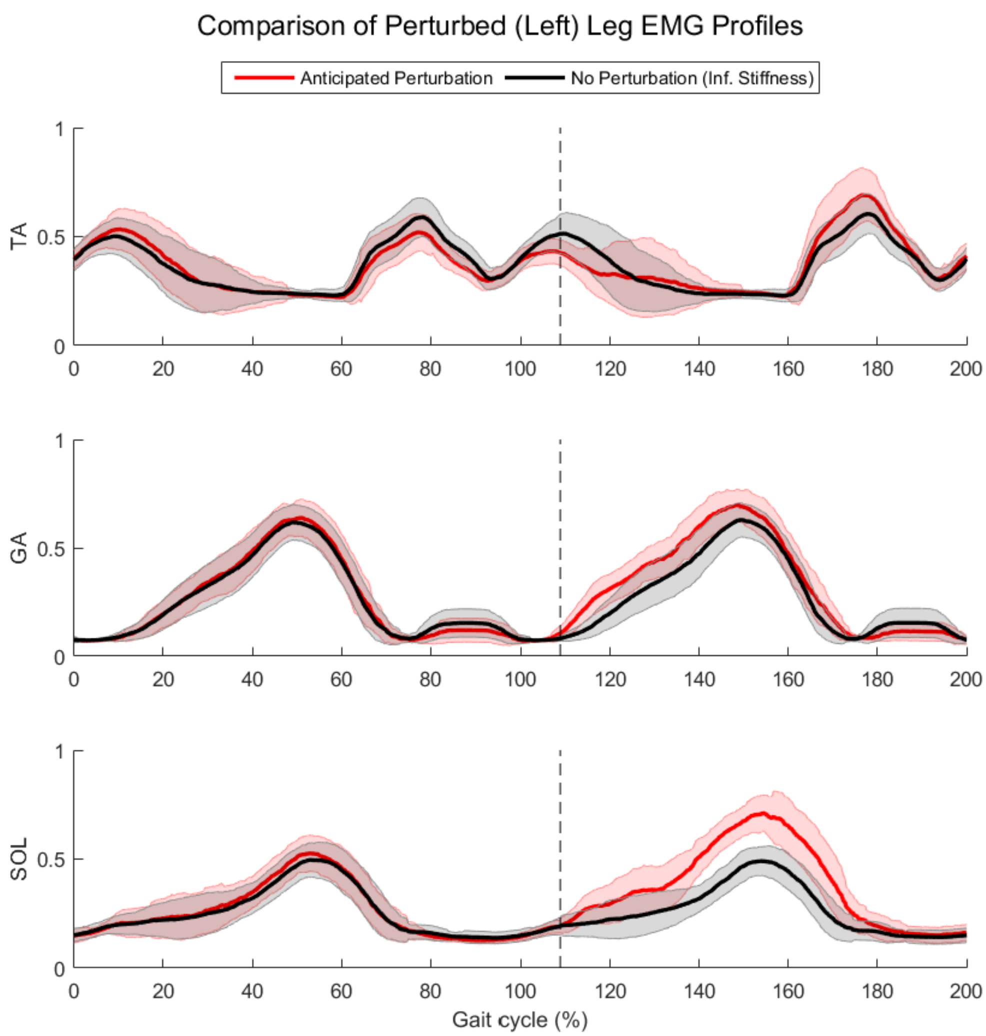

Figure 10: Comparison of Perturbed Leg EMG Profiles.

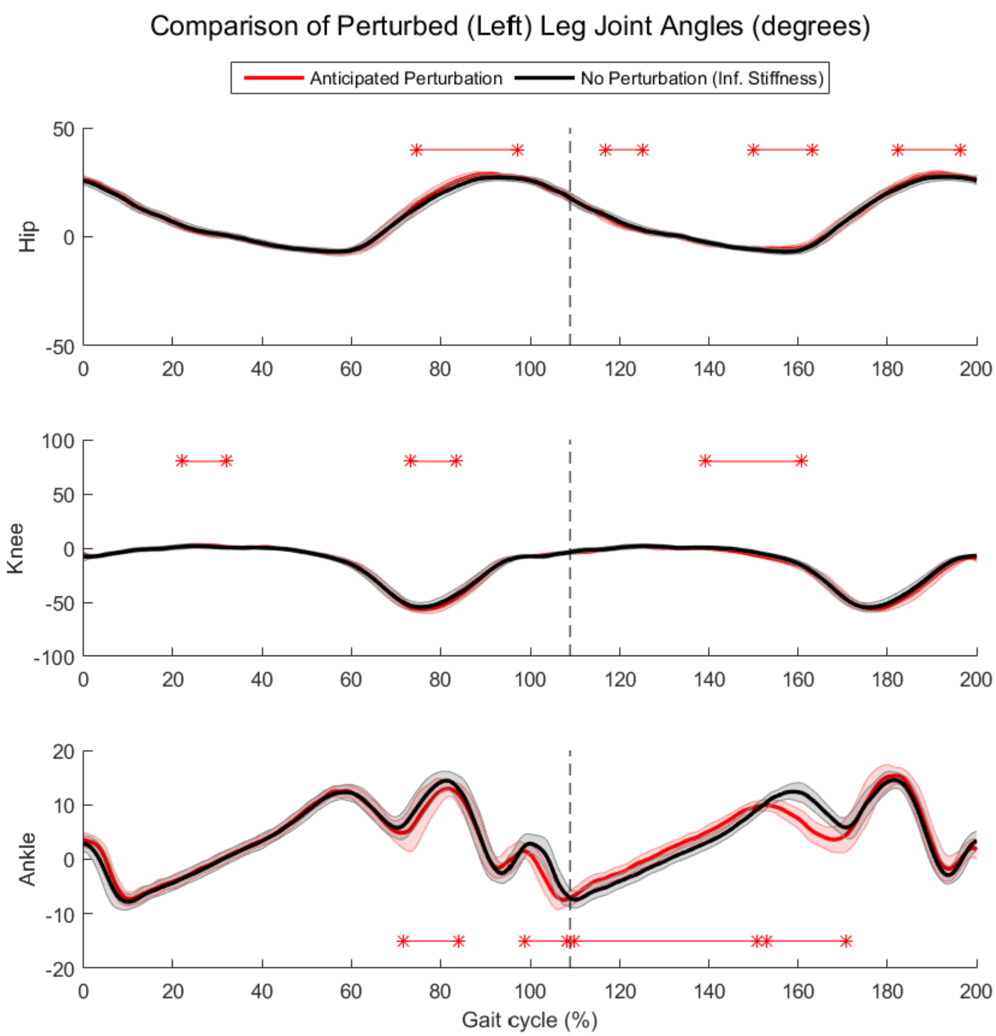

Figure 11: Comparison of Perturbed Leg Joint Angles.

It is also useful to examine kinematic data and explore another avenue that could indicate an anticipated walking surface change. Indeed, there are statistically significant differences in leg kinematics as depicted in Figure 11, which gives the comparison in joint angles between the perturbed cycles and the normal, rigid ground walking cycles. Changes in kinematics may be expected during a perturbation, but these results indicate that some of those changes also occur prior to the perturbation as well. For instance, there is a significant increase in hip angle during the swing phase of the perturbed gait cycle, which is seen in the figure in the 180-200% range. This is reasonably expected since this increase in flexion leads to increased toe clearance as the leg is brought up higher, which is a common response to encountering a compliant surface. Similarly, there is also an increase in hip angle during the swing phase before the perturbation even occurs, which is seen at 80-100% gait cycle. This result may suggest that not only does the subject react to the perturbed surface by adjusting their hip angle in response to a perturbation, but they also prepare for the change in surface stiffness by increasing hip angle during the swing phase before encountering the perturbed surface. There are also significant decreases in ankle flexion immediately before the perturbation, near 70-85% gait cycle and 100-110% gait cycle. Results in the ankle flexion of the perturbed leg do not contribute much to the control of a powered prosthetic since they could not be recreated by a transtibial amputee; however, these results do correspond to the qualitatively observed decreases in TA activation from Figure 10, which could possibly be replicated by a disabled person using their residual limb muscles. While there were no calculated statistically significant differences in TA activation, the evidence provided by the decreased ankle flexion from kinematics analysis supports a change in EMG profile and calls for further investigation in muscle activation.

These trends in kinematics seen in the previously mentioned representative subject are also observed in a second subject, which is demonstrated in Figure 12. There are statistically significant increases in hip angle at the end of the swing phase both before and during the perturbation, around 100-110% and 170-200% gait cycle, respectively. This consistency suggests a strong likelihood that hip angle could be a reliable indicator of anticipated surface change. Additionally, there is significantly decreased ankle flexion immediately before the perturbation, occurring at approximately 90-105% gait cycle. Again, this may not be directly useful for a powered ankle prosthetic but does support further investigation of EMG profiles since this trend is likely linked to a decrease in TA activation.

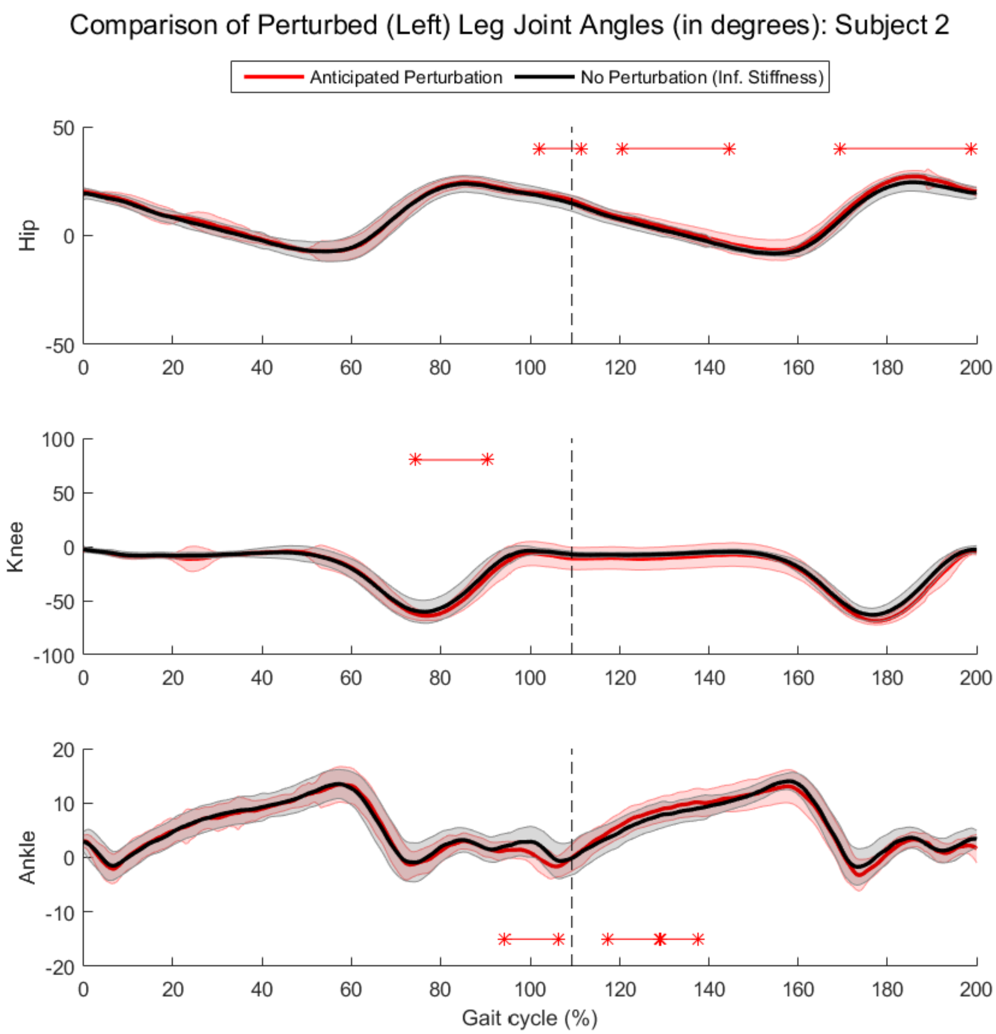

Figure 12: Comparison of Perturbed Leg Joint Angles in Additional Subject.

The changes in gait around the ankle observed during the transition between a level, stable surface and the compliant surface may be a product of walking on the VST in the same way that an individual's gait may differ from normal overground walking and walking on a conventional treadmill. Specifically, when walking on the VST, individuals tend to plant their foot on the treadmill's surface sooner after heel-strike compared to overground walking. This causes the trends in ankle flexion and TA activation seen in Figures 10-12 in the 100-120% gait cycle range in both walking modes. However, it is possible that this behavior is amplified when a perturbation occurs as individuals work to stabilize themselves, which explains why both TA activation and ankle flexion decrease immediately before and during an anticipated perturbation occurs.

4.1.2 Effects on Unperturbed Leg Muscle Activation and Kinematics

Additionally, it is possible that indicators of an expected surface change in kinematic and EMG data can be identified in the contralateral, or unperturbed leg through interlimb coordination. These results are unlikely to be useful in controlling a powered prosthetic since it would be inconvenient to require sensors on the unaffected leg, but they may strengthen the understanding of gait mechanisms and interleg coordination during walking surface transitions. Furthermore, the results could be relevant in studying the effects of perturbing the healthy, unaffected leg first.

Figure 13 shows a comparison of EMG profiles for the TA and GA of the unperturbed leg of a representative subject. At a glance, it appears there may be some evoked GA activation immediately before the perturbation affects the left leg, near 90-110% gait cycle. However, there is also some increased GA activity in the same location of the previous gait cycle, which can be observed close to 0% gait cycle. Seeing this behavior so far ahead of the perturbation suggests that this result is not unique to the gait cycles immediately before a stiffness change. Moreover, as was the case for the perturbed leg, statistical analysis does not indicate any significant differences between the two walking modes, possibly resulting from the naturally large variances in EMG during walking.

Kinematic data for the unperturbed, contralateral leg of a representative subject is given in Figure 14. Like the results from the perturbed leg, there is significantly increased hip flexion during the swing phase of this leg before and after the perturbation, which is found at approximately 40-60% and 120-150% gait cycles in the figure above.

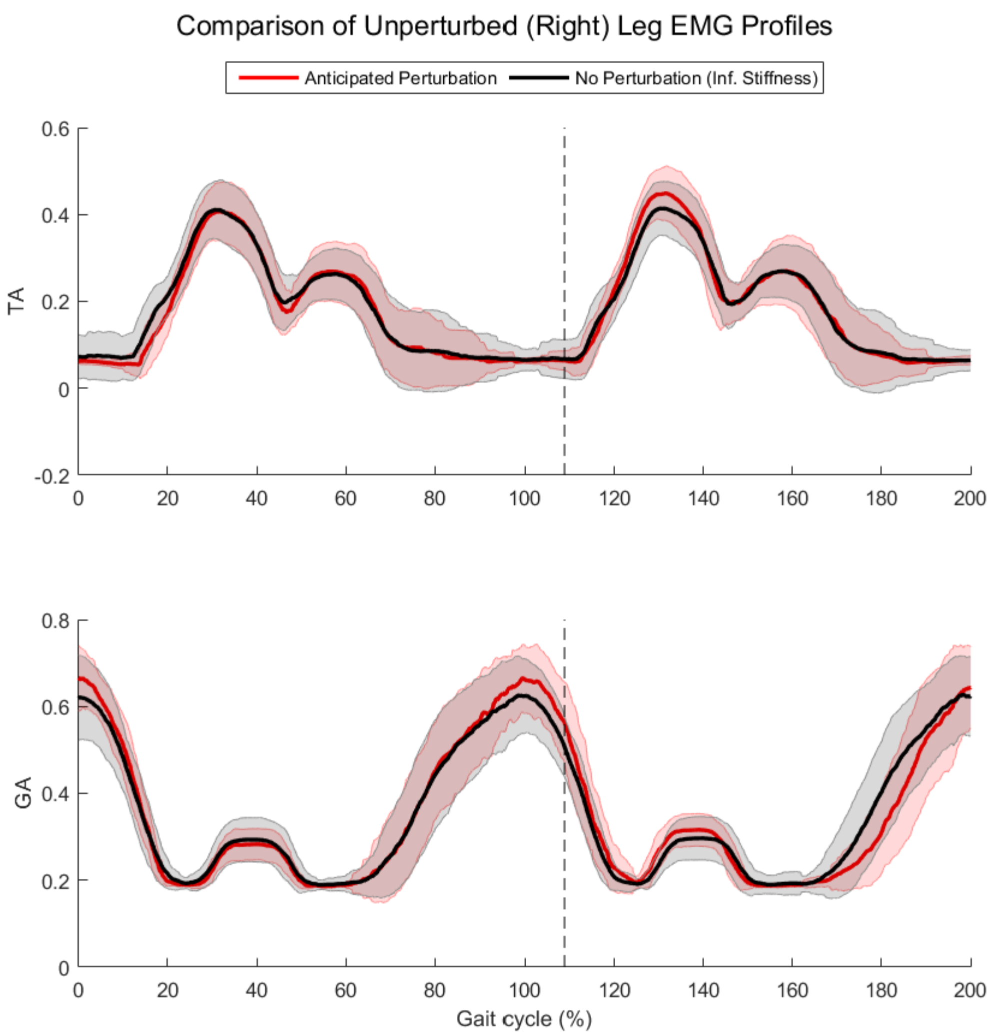

Figure 13: Comparison of Unperturbed Leg EMG Profiles.

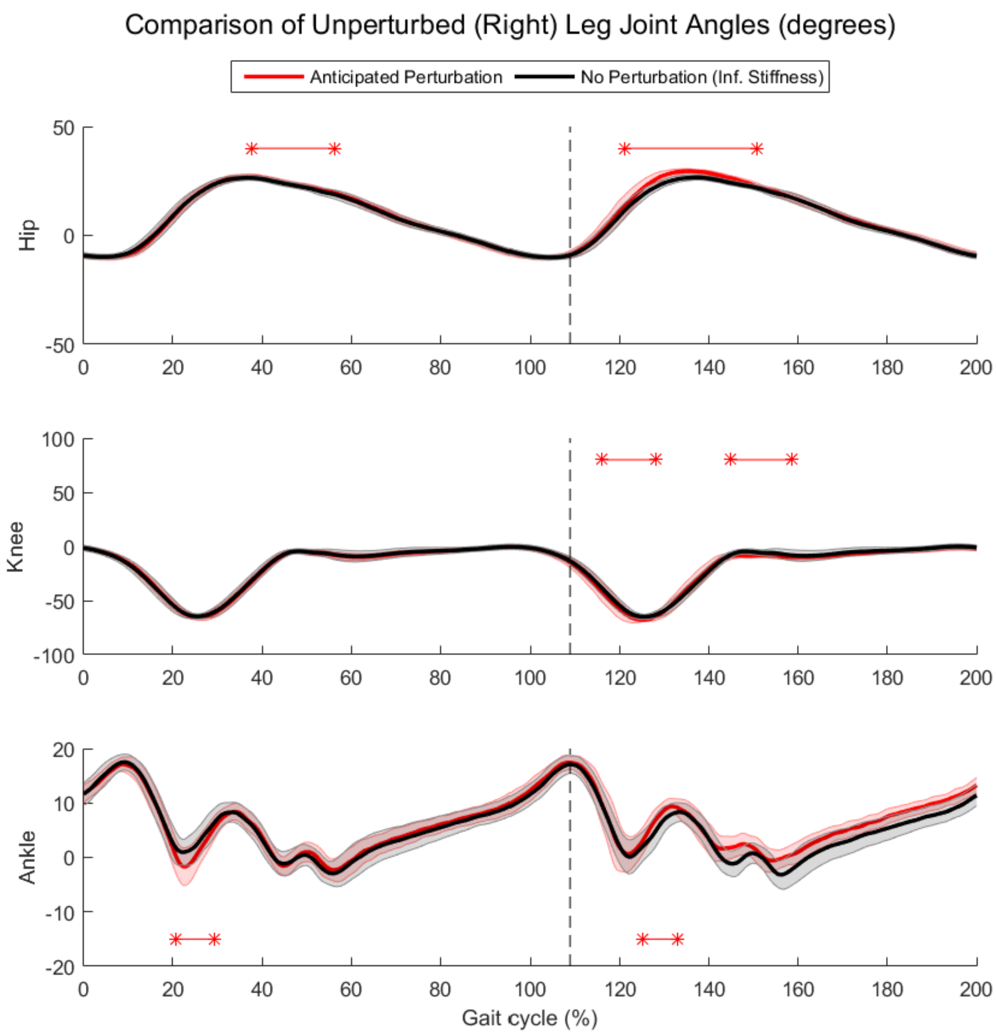

Figure 14: Comparison of Unperturbed Leg Joint Angles.

4.2 Experiment Limitations

The objective of the experiment was to search for and identify potential indicators of an anticipated walking surface change during a person's natural gait, particularly in the perturbed leg EMG profiles. The hypothesis theorized that changes in EMG are naturally made by individuals prior to the change in walking surface to ensure balance and normal gait patterns are maintained. The experimental setup attempted to recreate walking surface changes found in real-life in environments using the VST. Still, there are limitations of the experiment and its setup, which will now be addressed.

While the controlled environment of the VST provides useful advantages, there are also some drawbacks in using the system. Parameters such as walking speed are fixed, and only one leg encounters the perturbed surface while the other continues to contact level, rigid ground. These are elements that do not reflect normal walking, but the benefits of using the VST was enough to counter these effects. Still, it would be useful to design an experiment that captures the best features of the VST and maintain elements of natural walking. One of the more useful advantages of the VST was that its motion capture system allowed the data to be easily separated into individual gait cycles and by walking modes. However, other studies have successfully used wireless sensors such as accelerometers and load cells to determine instances of heel strike, which is then used to analyze data similar to how the motion capture system is used in this experiment.

Being able to perform experiments outside of the VST would also allow the study of gait on real-world surfaces. One conclusion from Paysant et al. was that gait does not necessarily change when the walking surface is changed (2006, 156). Rather, the characteristics of the walking surface must also change to expect significant differences in gait. During the experiment, the stiffness mechanism was commanded to a position during perturbation such that the effective surface stiffness range was dramatically low (20-35 kN/m) to create a stark contrast between walking on level ground and on a low-stiffness, compliant surface. While there are certainly surfaces with similarly low stiffness, it would be useful to specifically identify such real-world surfaces and confirm the VST's ability to simulate them. One method, for example, would be to measure the surface characteristics such as the stiffness and damping of real surfaces and simulate these parameters on the VST. Alternatively, since the research is primarily focused on gait, the natural gait of persons walking on different real-world surfaces could be studied and compared to the gait observed on the treadmill. For instance, if the joint angles and EMG profiles of a person on walking on loose dirt and walking on the VST at a given stiffness are proven to be the same, then it may be possible to conclude that the VST was successful at simulating that surface. Then, the results from VST experiments will be more credible in applying the findings to real-world applications.

Finally, the one of the most challenging obstacles in conducting this study was working with the small sample of data collected. Although the comparison of EMG profiles shown in Figure 10 qualitatively appeared to show some differences in walking during a surface transition and normal, rigid ground walking, statistical analysis did not indicate any quantitative significant changes between the walking modes. One reason for this is the relatively large variance in EMG compared to, for instance, kinematics. The variance may decrease if the number of samples during an experiment increases, which would increase the confidence of the test's ability to conclude statistical significance. This is exemplified in the statistical significance demonstrated in kinematic data: although the joint angles appear largely similar between the two walking modes, the variance is very small, so even a slight change in joint angle can be confidently concluded as statistically significant. In contrast, there was no such conclusions for EMG profiles, even when there was no overlap between the error bars and means. Generally, it would also be useful to expand the study to include more subjects as well. Data from the three subjects discussed provided some important insight into how gait is altered for an impending change in walking surface. There appears to consistent increases in hip flexion during the swing phases before and during a perturbation, and in both the perturbed and unperturbed legs, which could be useful for developing an adaptive powered prosthetic. In addition, there were significant decreases in ankle flexion immediately prior to a perturbation, which give evidence to a decrease in TA activation in the same portion of the gait cycle, which could not be proven to be statistically significant but certainly appears from observation to be true. Still, these results will need to be replicated across a larger, more diverse group of subjects to conclusively determine whether these trends are useable indicators of walking surface transition.

Chapter 5

CONCLUSION

The purpose of this work was to investigate gait mechanisms, specifically muscle activation data obtained using surface EMG, to test the hypothesis that there is a significant change in EMG profile during a transition from walking on a level, rigid surface to a softer or more compliant surface. From experimentation with three healthy individuals, the results do not indicate any significant change in EMG before or during an upcoming surface transition although, from examination of the data, there visually appear to be a difference in data. However, there were statistically significant changes in hip flexion in the swing phase prior to an anticipated perturbation in both the perturbed and nonperturbed legs, which is a potentially strong indicator of walking surface change identified by this work. Also, there were statistically significant decreases in ankle flexion in the perturbed leg that occurred immediately before the perturbation. This result is not useful by itself since this could not be replicated in an amputated individual for use with a powered prosthetic, but these changes in ankle flexion occurred at the same time TA activation appears to decrease. This change in muscle activation could not be statistically proven, but the evidence provided by the change in kinematics in unimpaired, healthy subjects supports the need to further investigate potential changes in EMG data. If such indicators could be identified, then future developments in adaptive protheses could utilize residual limb muscle activation to improve the walking ability in those with lower limb loss. Still, more research and continued experimentation should be conducted to more conclusively determine what, if any, indicators of surface change exist in EMG data.

In conclusion, this work cannot positively conclude that the hypothesis proposed at the beginning of the research, that the EMG profiles significantly differ between gait cycles during a walking surface change and normal, level ground walking, is true.

However, the findings determined from the kinematics analysis does show statistically significant indicators of surface transition and provide evidence that there may similarly be significant changes in EMG found in future experiments. This is important to the development of powered prostheses since it suggests that a person's natural gait, and the data that can be obtained from it, can potentially be used to control the prosthetic to adapt to a wider range of walking surfaces.

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