Anticipatory Muscle Responses for Transitioning Between Rigid Surface and Surfaces of Different Compliance: Towards Smart Ankle-foot Prostheses

by

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A Thesis Presented in Partial Fulfillment
of the Requirements for the Degree
Master of Science

Approved April 2019 by the
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May 2019

ABSTRACT

Locomotion is of prime importance in enabling human beings to effectively respond in space and time to meet different needs. Approximately 2 million Americans live with an amputation with most of those amputations being of the lower limbs. To advance current state-of-the-art lower limb prosthetic devices, it is necessary to adapt performance at a level of intelligence seen in human walking. As such, this thesis focuses on the mechanisms involved during human walking, while transitioning from rigid to compliant surfaces such as from pavement to sand, grass or granular media. Utilizing a unique tool, the Variable Stiffness Treadmill (VST), as the platform for human walking, rigid to compliant surface transitions are simulated. The analysis of muscular activation during the transition from rigid to different compliant surfaces reveals specific anticipatory muscle activation that precedes stepping on a compliant surface. There is also an indication of varying responses for different surface stiffness levels. This response is observed across subjects. Results obtained are novel and useful in establishing a framework for implementing control algorithm parameters to improve powered ankle prosthesis. With this, it is possible for the prosthesis to adapt to a new surface and therefore resulting in a more robust smart powered lower limb prosthesis.

ACKNOWLEDGEMENTS

I would like to express my profound gratitude first and foremost to my committee chair, Professor Panagiotis Artemiadis, for giving me the opportunity to carry out this research, who in attitude and substance stands for everything a great supervisor should be; his dedication and commitment in all aspects, his passion for excellence, and his value for time. Without his guidance, advice and support, this thesis would not have been a possibility. Additionally, I would like to extend my appreciation to my committee members, Dr. Marco Santello and Dr. Hyunglae Lee, for their interest in my work and for agreeing to play such an important role as members of my committee. To my lab partner as well, Emiliano Jose Quinones Yumbla who all throughout, provided me with the support and advice I needed to pull this through. Also, to all members of the Human-Oriented Robotics Control laboratory, thank you for being of great help as well. Thank you Mastercard foundation for giving me the opportunity to pursue higher education in such a prestigious university. Most importantly, I would like to thank God Almighty for giving me the grace to do this.

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Chapter 1

INTRODUCTION

1.1 Overview and Motivation

Locomotion is of prime importance in enabling human beings to effectively respond in space and time to meet different needs. It is the displacement of the entire body from one place to another. It should be noted that locomotion involves various locomotive movements such as walking, jumping, running, crawling, swimming etc. Locomotion is successfully achieved in humans by the interaction and movement of tissues and joints such as muscles, bone, cartilage, ligaments, tendons etc. Human bipedal locomotion has been shown to depend less on reflexive automaticity as in the case of quadrupeds[1] and more on integration with higher controls.

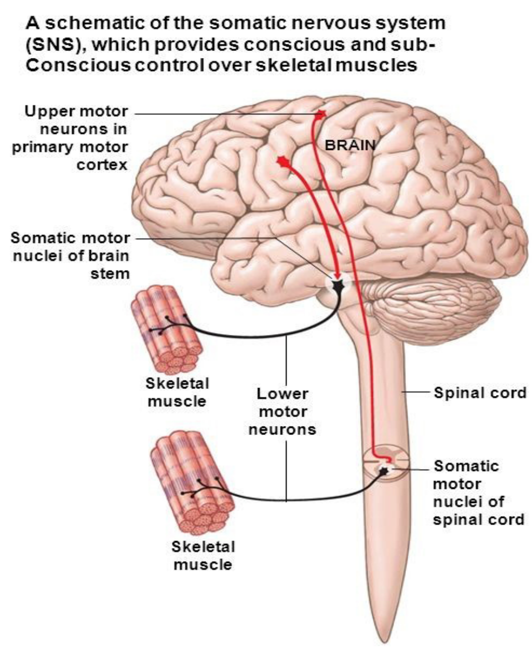

Figure 1.1: Skeletal Muscle Control System.

These controls involve the transmission of signals from the brain, via the spinal cord through a complex somatic motoneuron network as depicted in Figure 1.1 [2]. Walking, which is the effect of skeletal muscle activity, is controlled by different systems, one of which is the vestibular system, which collects information relevant for balance and control of movement[3]. Muscle activity to a great extent, sums up the bodys response to changes in the environment while walking. However, for some individuals, due to limb loss being one of the factors, this entire process has been tampered with and walking is a great challenge.

In the US, approximately 2 million individuals live with limb loss with an estimated increase of about 3.6 million by the year 2050[4]. Of these, amputations of the lower-limb are the majority, representing approximately 71% of dysvascular amputations amputations occurring from the poor vascular status of the limb[5, 6].

Amputation of one or both lower limbs poses long term physical and psychological challenges for amputees with major issues relating to balance, falling and the fear of falling[7]. After an amputation, such individuals depend on the prosthetic devices they are provided with to return to their normal lives and daily activities. Regrettably, these devices are more suited for walking on level ground and users experience shortfalls when complicated walking conditions are encountered. Figure 1.2 depicts the gap of current prosthetic devices in achieving high human volitional control as in the case of natural human walking. Standard prosthetic limbs are capable of restoring walking capabilities but are yet to replicate natural walking in more complicated walking conditions.

Research indicates that approximately 52% of out-patients fall with major reasons related to the prosthesis they use[8]. With the ankle joint being the most critical joint for gait stability and propulsion [9], extensive work has been done regarding human gait to improve the design of powered ankle prostheses in dynamic walking conditions[10, 11, 12, 13, 14]. A very important aspect of walking is an adaptation to terrain, and although previous studies have been successful in utilizing control strategies for walking and running with powered ankle prosthesis[15, 16, 17], there is a loophole in adapting to compliant surface walking. To advance current state-of-the-art lower limb prosthetic devices, it is necessary to adapt performance at a level of intelligence seen in human walking. Hence, an understanding of how able-bodied humans integrate sensorimotor control mechanisms resulting in robust gait control in dynamic walking is highly relevant. Limited joint angle mobility at the prosthetic end of lower limb amputees and the lack of distal muscles and sensory feedback from the lower limb results in difficulties while walking on uneven or non-rigid surfaces. Young, active trans-tibial amputees have been shown to increase toe clearance by increasing hip and knee flexion on the prosthetic side while increasing knee and ankle flexion on the intact limb during locomotion on a destabilizing rock surface[12]. Furthermore, Gates' study disclosed that variability of all step parameters and kinematic measures are affected by the surface type[11]. Intact individuals take conservative measures such as increasing minimum toe clearance to improve stability on complex

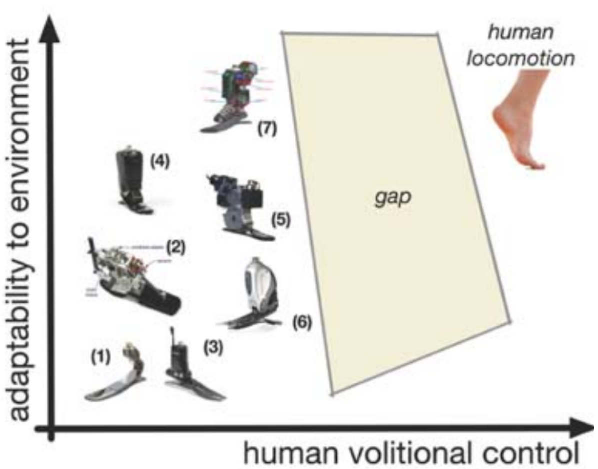

Figure 1.2: Available Prosthesis and Gap in Human Volitional Control and Environment Adaptability.

surfaces and reduce the likelihood of falls. In addition, it has been demonstrated that there is a shift in the synchronization of muscle activation while walking on cross-sloped surface compared to level-ground walking. This reveals the existence of a possible feed-forward system for control as small alterations to the walking surface were demonstrated to have significantly altered gait patterns[13]. Studies done in the past with human runners proved that stiffness of stance was adjusted to accommodate surface stiffness during steady-state running[17]. However, there was no indication of anticipatory response of specific muscles prior to transitioning to the compliant surface. While those previous studies are useful in understanding gait mechanisms involved in walking over some common obstacles, they were limited to hard, rigid surfaces, which only encompasses a limited type of natural environment individuals encounter daily. Additionally, studies carried out involving compliant surfaces identified the muscle activity only during walking on the compliant surface. Despite the progress made in research findings, a gap remains in the ability of amputees using powered ankle prosthesis for the maintenance of balance and stability when traversing complex, and especially compliant, terrains.

1.2 Proposed Work

This thesis is focused on the mechanisms involved during human locomotion while transitioning from rigid to different compliant surfaces. I hypothesize that; 1.) There is statistically significant difference in muscle activation just before and right after stepping on a compliant surface and 2.) There is variation in muscle activation for different compliant surfaces. The results observed can be used in the control of advanced powered ankle-foot prostheses to eventually achieve natural and robust walking on compliant surfaces for lower limb amputees. Figure 1.3 indicates how electromyographic(EMG) data would be used in control of prosthesis by providing

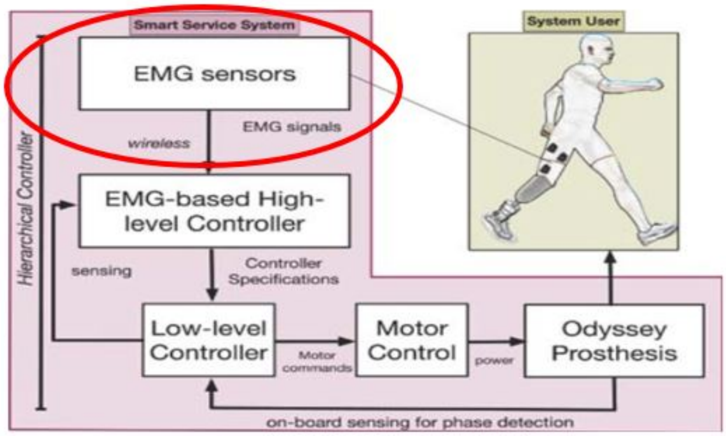

Figure 1.3: Architecture of Powered Ankle Prosthesis Control System. Indicating EMG Signal use in Control

real-time information. For a more robust system, a better understanding of muscle behavior is relevant; knowing which specific muscle and how that muscle responds would provide a good foundation on how to replicate this behavior for non-intact subjects, in this case, for lower-limb amputees.

In this work, a unique tool, the Variable Stiffness Treadmill (VST), is utilized as the platform for human locomotion, simulating rigid to compliant surface transitions. The ability to simulate transitions between rigid and non-rigid surfaces, while measuring lower limb muscle responses, creates a window on sensorimotor control strategies for dynamic walking that has not been explored previously. The results of this study indicate solid evidence that when human subjects are prepared to transition from rigid to compliant surfaces, there exist significant muscle activity alteration that precedes the step onto the compliant surface. Additionally, the results indicate that muscle alterations vary depending on the surface type. An application of these result to vary control parameters for powered ankle prosthesis to properly adapt to a compliant surface will increase gait performance and stability for amputees.

1.3 Thesis Organization

The rest of this thesis is organized as follows: Chapter 2 discusses relevant literature that is useful in understanding similar work that has been done in the past, specifically, muscle activity and kinematic strategies while walking on different surfaces and further addresses the gap and limitations that exist. Chapter 3 provides information on all experimental methods as far as the setup of the system, experiment protocol and data collection and analysis for EMG data collected while subject walked on the Variable Stiffness Treadmill(VST). Chapter 4 expands on the analysis of EMG activation results obtained for the different compliant surfaces used in this study. Finally, Chapter 5 concludes the thesis with a discussion and recommendation for future work included.

Chapter 2

LITERATURE REVIEW

2.1 Muscular Activity and Synchronization on Different Surfaces for Prosthetic Device Control

Muscles are controlled by the nervous system and the output from the nervous system to the muscles is based on the stimulus received. The goal of this thesis is to identify muscle responses while intact humans transition during walking from one compliant surface type to another. Hence, this section expands upon published work relating to muscle responses as well as general gait adaptations in response to varying walking surfaces.

Estermann et al. investigated muscular synchronization between walking on flat and cross-sloped surfaces and introduced the fact that movement and muscular adaptations are necessary for daily living. This is generally expected as different types of stimulus will generate different types of responses. Also, based on the type of stimulus, different muscles will be activated, and the duration and amplitude of the activation are also expected to vary. The focus was drawn on the evidence of anticipation in muscular activation patterns in performing certain tasks in a specified manner. More specifically, Estermann et al. analyzed the extent of the temporal shift in muscular synchronization while walking on a flat and cross-sloped surface. Since most biomedical signals are functions of time [18],it is ,therefore, necessary to understand the fundamentals of such signals as they relate to time before further inferential analysis are made.

In the study carried out by Estermann et al., a total of nine muscles of the right leg were investigated for walking on a flat surface and a cross-sloped surface inclined at an angle of about 6 degrees. Specifically, the muscles were the Tibialis anterior (TA), Peroneus longus (PL), Gastrocnemius medialis (GM), Gastrocnemius lateralis (GL), Vastus medialis (VM), Vastus lateralis (VL), Rectus femoris (RF), Semitendinosus (ST) and the Biceps femoris. This provided an indication of the types of muscles to investigate for the purpose of this work. With a focus on what happens just before and after stepping on a compliant surface from a rigid terrain, there was the need to investigate lower limb muscles that are activated during the swing phase and just after heel strike. These are; 1. The TA - responsible for dorsiflexion of the ankle joint at heel strike and 2. the GA, PL and, Soleus (SOL) - responsible for plantar flexion of the ankle joint during the swing phase of the gait cycle. Per the results, a significant difference was observed for knee flexion and ankle dorsiflexion during the stance phase.

Though only rigid surfaces were investigated, an instance of the surface -surface condition is addressed by comparing level ground walking to an inclined surface. The conclusion drawn is that even with small changes in surface condition, there is a detectable change in the kinematic and muscular activation pattern [13]. One interesting observation in the experiment is the fact that subjects were made to walk barefooted. The reasoning behind this is not explained in detail. However, though the usefulness will most probably be to get enough sensory information, most everyday walking is done with footwear on.

Furthermore, Huang et al. undertook an interesting study which will be of significance to the long-term application of this thesis. In their study, they developed an algorithm based on neuromuscular-mechanical fusion to continuously recognize a variety of locomotion modes performed by patients with transfemoral amputations. EMG signals were recorded from the gluteal and residual thigh muscles and ground reaction forces/moments measured from the pylon were used as inputs to a phase dependent pattern classifier for continuous locomotion-mode identification. The results obtained showed that neuromuscular-mechanical fusion outperformed methods that used only EMG signals or only mechanical information.

With the development of intent recognition strategies, there is an intuitive prosthetic leg control. One of these methods is known as the "Echo control" where the prosthetic leg repeats the motion of the sound leg. While this work does not involve directly, the control strategies used, this piece was found very useful in understanding the role of the contralateral leg. With regards to experiments carried out in this work, perturbations were applied only to one limb. However, muscle activation was recorded for both limbs. In their article, the assumption used in "echo control" that the motion of the two legs is symmetric during locomotion was argued out. Due to the fact that motions of lower limbs are different in the transition from level-ground walking to stepping over an obstacle[19], the assumption of symmetry is not always valid. This is probably one of the reasons why current prosthesis that uses this approach fail and about 50% of users tend to fall as a result[8]. From this study as well, it was made known, that while mechanical sensors respond to the patients movements, EMG signals precede movement onset and may be used to help predict task transition. This aligns very well with the purpose of this thesis, to identify anticipatory muscular responses and implement this using good control strategies for ankle-foot prostheses control. Reflecting on the study on muscular synchronization, in their discussion of the results, they concluded that there was some sort of feed-forward system controlling the response[13], an anticipated muscular synchronization which agrees with the conclusions of Huang et al.

Additionally, an investigation of muscle activity and movement variability relative to the center of pressure (COP) indicated a significant cross-correlation between COP and the ankle joint as well the peroneus longus muscle during a unipedal stance on a solid and compliant surface[10]. Compliant surfaces tend to disturb upright stance by reducing sensory input and decreasing effective use of ankle torque. The decrease in ankle torque can be most associated with reduced muscular activation on the compliant surface. This study focused more on stability during a unipedal stance on foam and on an air-filled disc, investigating control strategies for maintaining stability. As in the case of this thesis, EMG data was collected from the TA, GA, SOL, and PL among other muscles. Force data was collected using force plates and the anterior-posterior and mediolateral center-of-pressure traces were calculated as well. For analysis, they determined the mean cross-correlational curves for each subject and for each of the surfaces.

From Figure 2.1, it can be inferred that the ankle dorsiflexion angle and eversion angle have a very high correlation with the center of pressure in the anterior-posterior and in the mediolateral direction respectively. This affirms the fact that the ankle joint is the most critical joint for gait stability and balance [9]. The end goal of this thesis is to improve balance and stability in amputees. Hence, narrowing on ankle joint dorsiflexors and plantar flexors.

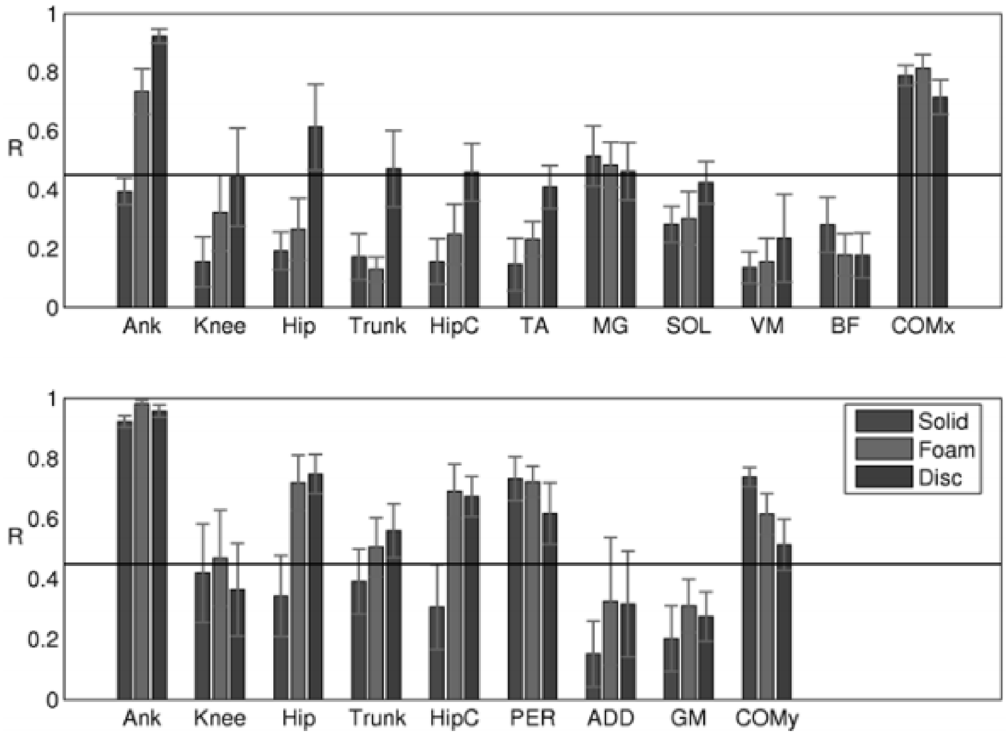

Figure 2.1: Means (SD) of Maxima of Cross-correlations with Center of Pressure. Horizontal line level of significance. R raw correlation values. Upper chart correlations with the anterior-posterior component of the center of pressure. Ank = ankle dorsiflexion angle; Hip = hip flexion angle; Trunk = trunk extension angle; MG = medial gastrocnemius activity; COM = anterior-posterior location of center of mass. Lower: maxima of cross-correlations with the mediolateral component of the center of pressure. Ank = ankle eversion angle; Knee = knee abduction angle; Hip = hip abduction angle; Trunk = trunk L side flexion angle; HipC = contralateral hip adduction angle; PER = peroneus longus (fibularis longus) activity; ADD = adductor activity; GM = gluteus medius activity; COM = mediolateral position of center of mass. (Adapted from Croft et al.)

2.2 Kinematic Strategies and Other Locomotion Modes on Different Surfaces.

Previous studies indicate the existence of an inherent relationship between muscle parameters and kinematic variables[20, 21]. There have been instances of high correlation between specific muscle kinetics and kinematics as associated with specified movements both in the cases of able-bodied and impaired individuals. As such, it is, therefore, necessary to understand the existing knowledge of kinematics on compliant surfaces. Hence, this section discusses previous research investigating gait kinematics in response to different surfaces.

Gates et al. investigated the kinematic strategies employed for walking on a destabilizing rock surface. Most outdoor falls are said to be caused by uneven surfaces[22] and these surfaces are known to apply both mechanical and kinesthetic perturbations. Like in this thesis, the focus is drawn on mechanisms adapted when there are perturbations of some sort. A lower center of mass (COM) may enhance stability by decreasing the moment arm between the COM and the ground reaction force such that a greater amount of force is needed to induce a fall. Their study quantified lower limb joint kinematics, center of mass height (COMVT) and minimum toe clearance (MTC) while subjects walked across a level ground (LG) and on a destabilizing loose rock surface (RS) as depicted in Figure 2.2 at four control levels of speed utilizing Froude number for normalizing walking speed of subjects.

Toe clearance, which is the distance between the foot and the ground when the foot swings forward during normal walking could give an indication of the behavior of muscle activity more specifically during the swing phase before the leg contacts the ground again. In their work, it was observed that at faster speeds, subjects increased step length (SL) and decreased step time (ST) but maintained step width (SW). However, the difference in the average step parameters was not significant[11]. Additionally, the surface type was shown to affect the variability of all step parameters. More specifically, the variability in these parameters for walking on the RS was shown to have increased by two-fold as compared to LG as depicted in Figure 2.3. More importantly, the interaction effect for step length variability was significant. This variability is expected as the surface is not rigid. Furthermore, the MTC was found to be 3.8 times greater on the RS as compared to LG and MTC increased with speed only on the RS as depicted in Figure 2.4 section B. This can be most likely

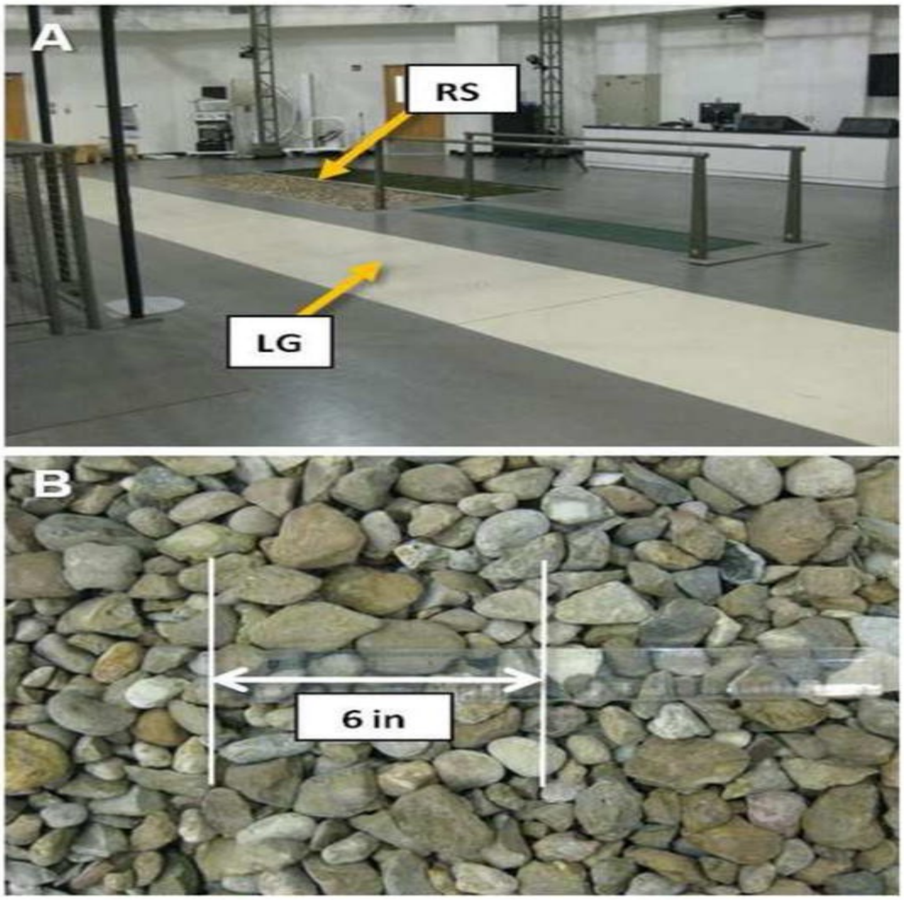

Figure 2.2: Experimental Setup by Gates et al. Depicting Level Ground and Destabilizing Rock Surface

associated with the increase in hip and knee flexion and ankle dorsiflexion during the swing phase on the rock surface. This is very insightful and useful to know while critically investigating the muscle responses on the different compliant surfaces.

kinematic strategies for walking on compliant surfaces has been investigated. However, there has been no focus on response during the transition period. In Gates study, the surface types were analyzed separately without considering transitioning from one surface to another. This thesis delves into muscle responses for transitioning between different surfaces and as muscle activation is known to precede movement, it can serve as an indication of the type of kinematic response to expect while transitioning.

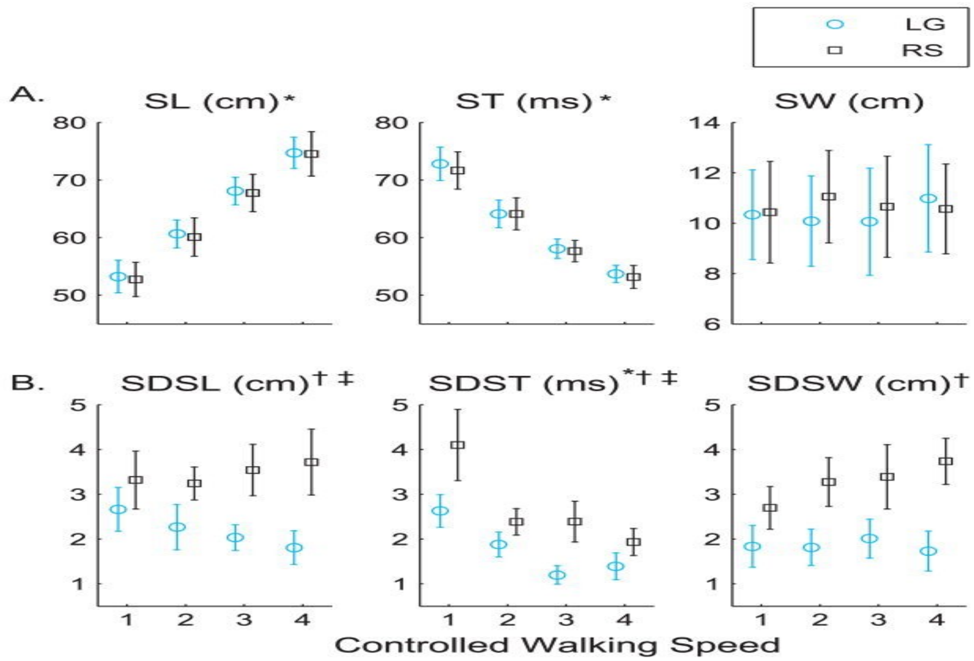

Figure 2.3: Temporal Distance Parameters from Gates et. al. A) RS Rock surface LG Level ground SL -Step length ST Step time SW- Step width Vertical axis walking speed B) Average within-subject variability of SL, ST and SW

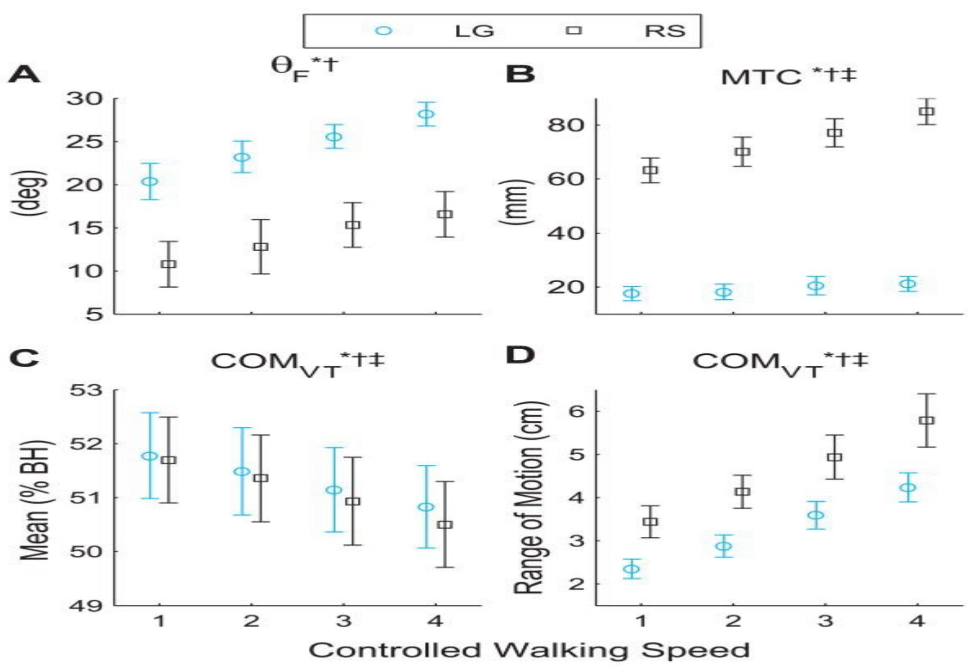

Figure 2.4: B. Minimum Toe Clearance for both Destabilizing Rock Surface and Level Ground

Chapter 3

METHODOLOGY

This section provides insight into the rationale and techniques employed in carrying out this experimental research. As this research investigates muscle responses prior to and right after stepping on compliant surfaces, the first section explains the basis for the stiffness levels chosen to be simulated. The subsequent sections explain the various components of the experiment, the role they played and how they were put together to test the hypothesis and draw conclusions on results obtained.

3.1 Surface Stiffness Determination

With a focus on different surfaces of different stiffness levels, there was the need to simulate stiffness values that will clearly differentiate one surface type from another without presenting an uncomfortable experience to subjects. This section answers the basis upon which the VST was programmed to achieve the desired environmental simulations.

A few examples of compliant surfaces encountered on a regular day includes beach sand, gravel, grass among others. Different surface types will exhibit different stiffness properties. It is important to note that no surface type will have a single unique number representing its stiffness but rather the stiffness properties of the surface are described by a range of values. The stiffness of a surface in response to an axial load is given by the equation

$$k = \frac{P}{\delta} \tag{3.1}$$

Where P is the axial stress applied and δ is the deflection produced as a result.

Considering Figure 3.1, the behavior of a rigid surface such as pavement is de-

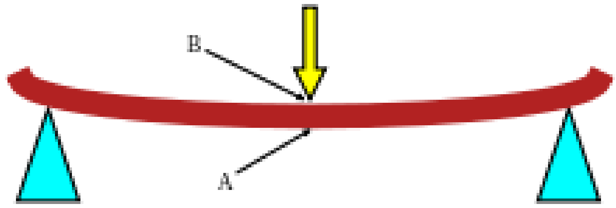

Figure 3.1: Response of a Surface to an Axial Load.

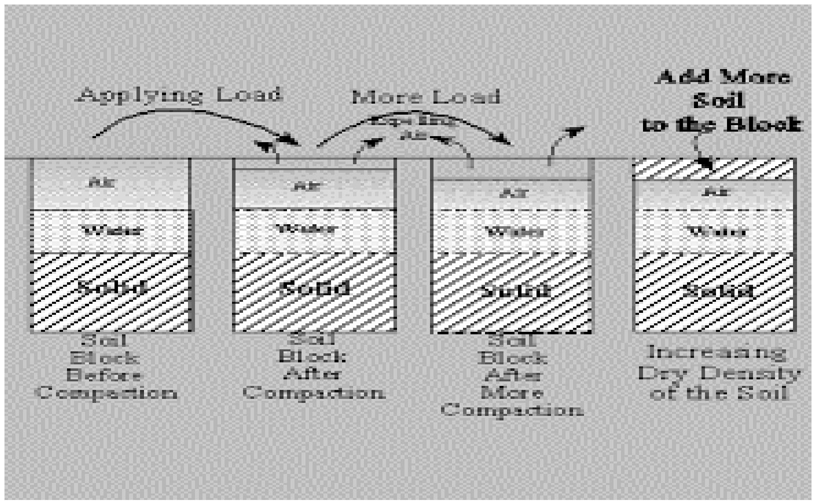

Figure 3.2: Soil Surface Response to Increasing Loads

picted. In this scenario, with the application of an axial load, B, there is a change in length of the material A. Rigid surfaces usually experience very minimal deflections in response to the forces exerted while walking. The second Figure, Figure 3.2 on the other hand best depicts surfaces that are compliant. The compliance of an object is inversely related to its stiffness. With the application of a load, in this case, the force exerted by the foot, the amount of air present reduces. The more compliant a surface is, the less closely packed the individual particles that make up the surface area. Hence, an individual walking on surfaces of very high compliance tends to have their foot sinking more into that surface.

The Youngs modulus or modulus of elasticity can be related to stiffness although these two properties are different. The Youngs modulus of a material is a measure of the materials resistance to elastic deformation under load and it is an intensive property whereas stiffness is an extensive one. Typical values of Youngs Modulus for granular material[23] are indicated in Table 3.1 with units in MPa. Appropriate stiffness values to be implemented were obtained using these values.

Table 3.1: Typical Values of Young's Modulus for Granular Material

USCS	**Description**	**Loose**	**Medium**	**Dense**
GW,SW	Gravels/sand well-grained	30 - 80	80 - 160	160 - 320
SP	Sand, uniform	10 - 30	30 - 50	50 - 80
GM, SM	Sand/ Gravel Silty	7 - 12	12 - 20	20 - 30

CALCULATING STIFFNESS FROM YOUNGS MODULUS

$$YM = E = \frac{\sigma}{\epsilon} = \frac{F/A}{\delta L \ / \ Lo} = \frac{F.Lo}{A.\delta L} \tag{3.2}$$

$$\delta L = \frac{F.L}{A.E} \tag{3.3}$$

$$K = \frac{F}{F.L/A.E} = \frac{A.E}{L} \tag{3.4}$$

Equation 3.2 represents the formula for Young's modulus of a material which is given by the product of the force applied and the original length divided by the product of the area of surface and deformation. Equation 3.4 represents the formula for the stiffness of a material, K, which is the force divided by the deformation. Using these equations, the stiffness of a surface can be determined given the Youngs modulus of the surface. Based on previous literature[14], and knowing that the average step length of males and females is 0.71625 m, the assumed dimensions used for the compliant surface of the experiment were; 0.71625m long, 0.46m wide and 0.10m high. Assumption: A/L from equation 2 is a constant value C. Where C represents the compliant surface dimensions. Focusing on the surface area of the medium, the total surface area (TSA) of the compliant surface is given by the equation:

$$TSA = 2\ (\ lw + wh + hl) \tag{3.5}$$

$$= 2\ [(0.71625)(0.5) + (0.5)(0.10) + (0.10)(0.71625)] = 2\ (0.47975)$$

$$\therefore TSA = 0.9595m^2$$

$$C = \frac{TSA}{L} = \frac{0.9595m^2}{0.71625m} = 1.3396m$$

Using the values of the medium range from table 3.1 and with the assumption that the surface under consideration is of a much smaller area, the Youngs modulus values are scaled into KN/m2. The table below indicates the final stiffness values for three compliant surface levels.

$$Assumption : C_d = \frac{C}{10^3} \tag{3.6}$$

Table 3.2: Stiffness Values in KN/m used to Simulate Real-world Compliant Surfaces

USCS	**Description**	**Young's modulus**	**Stiffness**
GW,SW	Gravels/sand well-grained	80 - 160	107.68 - 214.336
SP	Sand, uniform	30 - 50	40.188 - 66.98
GM, SM	Sand/ Gravel Silty	12 - 20	16.0752 - 26.792

With this information, it was possible to program the VST such that the subjects could liken the feel of the perturbation to familiar surfaces. In the experiment, two stiffness levels were simulated and this will be explained further in the subsequent sections.

3.2 Experimental Setup

In this research, lower limb muscle activation in response to varying stiffness levels are investigated. The experimental setup is designed to simulate real-world walking on two surface types while recording muscle activation and joint kinematics of the subject.

Specific Aims: The goal of this experiment was to:

Identify and establish a fundamental relationship between the transition responses of the various lower limb muscles on one compliant surface type as considered to another.

It is expected that as the surface compliance differs, the muscle response will differ as well but how significant and consistent will this difference be across subjects. This

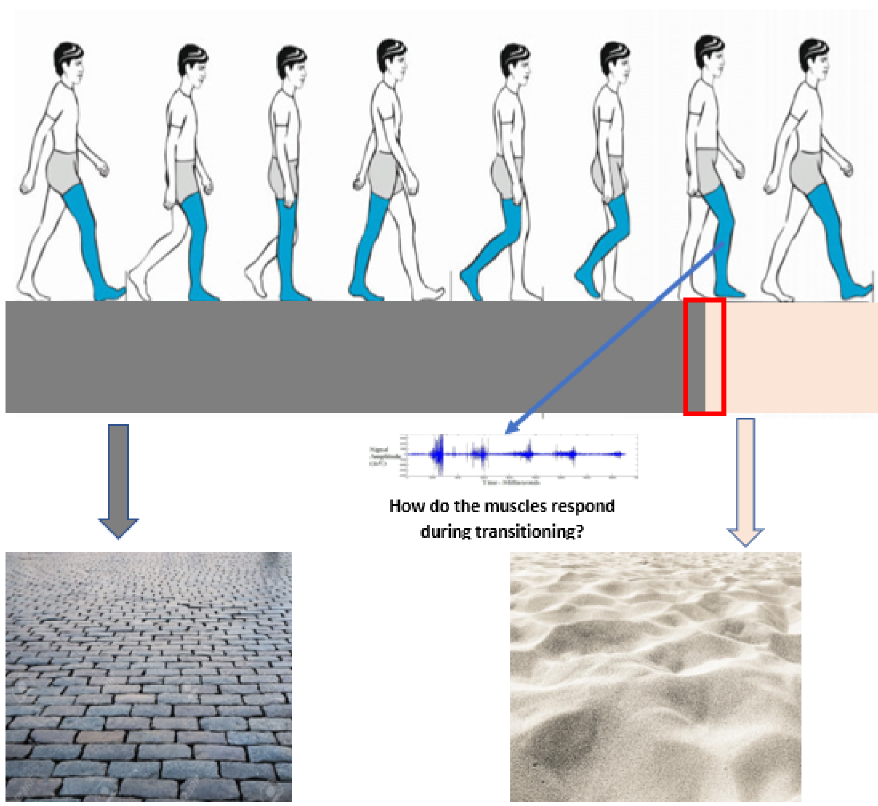

Figure 3.3: Real-world Big Picture Illustration of Experimental Setup.

is very important to consider in the control of an ankle prosthesis as this would be needed to determine what error range is allowable and how fine-tuned it should be. With the design of this experiment, it is necessary to develop a broad perspective of the transition relationship and based on results, further studies would need to be carried out to fine tune the system. Figure 3.3 depicts the big picture of what is simulated in the experiment.

Human walking transitions are simulated using a novel tool, the Variable Stiffness Treadmill developed by Skidmore, Barkan, and Artemiadis which will be expanded upon in the next subsection. During the experiment, unilateral stiffness perturbations are applied while the subject walks on the treadmill to investigate muscle responses.

3.2.1 The Variable Stiffness Treadmill

The variable stiffness treadmill, by regulating the walking surface stiffness in real time, has the capability of controlling the load feedback stimulus. It has a wide range

Figure 3.4: The Variable Stiffness Treadmill Developed by Skidmore, Barkan and Artemiadis

of controllable stiffness theoretically from zero to infinity while maintaining a high resolution. In addition, the compliance of the treadmill surface can be actively varied within the gait cycle. This special feature allows the flexibility of determining when perturbations should occur, their timing, duration, and stiffness levels. The ability to vary stiffness levels allowed different compliant surfaces to be created for the purpose of simulating the environment. The VST is shown in figure 3.4.

To change the effective stiffness of the treadmill, the stiffness mechanism on the track is re-positioned. Figure 3.5 depicts the stiffness mechanism employed in the VST. It utilizes a high-capacity linear track (Thomson Linear, Part Number: 2RE16-150537) and a precision drive (Kollmorgen, Part Number: AKD-P00606-NAEC-0000). The development and analysis of the mechanism are expanded upon by Skidmore, Barkan, and Artemiadis (2014, 1717-24).

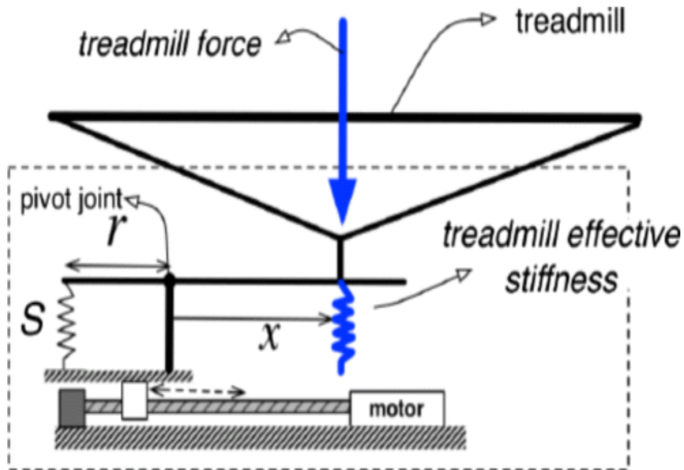

Figure 3.5: Stiffness Mechanism of the VST - Conceptual Diagram

The vertical stiffness of the walking surface is varied by controlling the kinetic and kinematic interaction between the subject and the walking surface. The stiffness mechanism consists of a spring-loaded lever mounted on a translational track. By design, the coefficient of stiffness, S of the linear spring and the moment arm, r through which it exerts its force remains constant. Varying the distance, x of the treadmill controls its effective stiffness. Hence, by changing the distance x, different surface levels are simulated. To accommodate the stiffness mechanism, the treadmill belts are supported 70cm above ground level to enable deflection of the belt. This mechanism is only employed to the left belt of the VST. Each belt of the VSTs effective stiffness can range from its minimum value (61.7N/m) to its maximum, which is theoretically infinite and this is used to implement rigid surface walking[24].

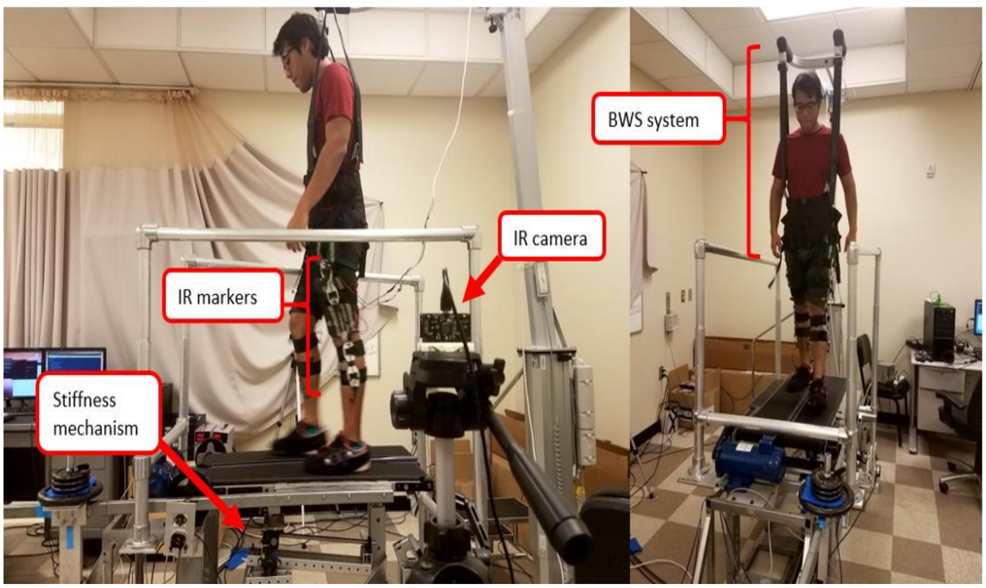

Figure 3.6: The VST Platform Experimental Setup. The infrared camera system for tracking the leg motion is shown (IR camera, IR markers), along with the body-weight support (BWS) that was used solely for subjects' safety during the experiment. The stiffness mechanism can alter the effective stiffness of the treadmill

The experimental set up also includes a motion capture system which consists of two infrared (IR) cameras positioned on the side of the treadmill as depicted in figure 3.6. The cameras are calibrated to capture IR marker positions as the subject walks on the treadmill. Additionally, the body weight support system (BWS) shown in the figure 3.6 was only used as a safety precaution during the experiment. The motion capture system is further described in section 3.4.1.

3.3 Experimental Protocol

This section describes exactly how perturbations were timed and allocated during the experiment to gather data necessary for evaluating the anticipatory behavior of the muscles. A total of 6 healthy subjects [age 21.5 $\pm$ 3.5 years, weight 166 $\pm$ 50 lbs] undertook this experiment. Each subject was made to walk on the VST for approximately 10 minutes per experimental block (a total of two blocks) and was verbally notified throughout the experiment three steps before a perturbation occurred. The perturbations were programmed to occur within a specific point in the gait cycle.

Subjects walked on the VST at a speed of 0.60m/s for a minimum of 380 gait cycles. The right belt only delivered infinite stiffness, representing a rigid surface, throughout the entire experiment. The left belt, on the other hand, was commanded to deliver a stiffness of 100KN/m for block one and 60KN/m for block two. These are the stiffness levels determined in section 3.1. The two blocks of the experiment run were the same except for the stiffness delivered by the treadmill based on the linear track position. Moving the linear track to 4cm and 5.3cm resulted in an effective stiffness of 100KN/m and 60KN/m respectively. The break between the two blocks was to allow the subject to take a break and to ensure that the EMG sensors and the IR markers placed on the subject had not shifted in position as this would affect the quality of data collected either being unrepresentative of the measure or no data being collected at all.

Figure 3.7 depicts the experimental flow of the treadmill motion as well as the duration and implementation of stiffness in the VST as the subject walked on the treadmill. The experiment initiates with the subject walking on a rigid surface (K = 1MN/m) for 30 cycles. This is used to establish a reference profile for rigid surface

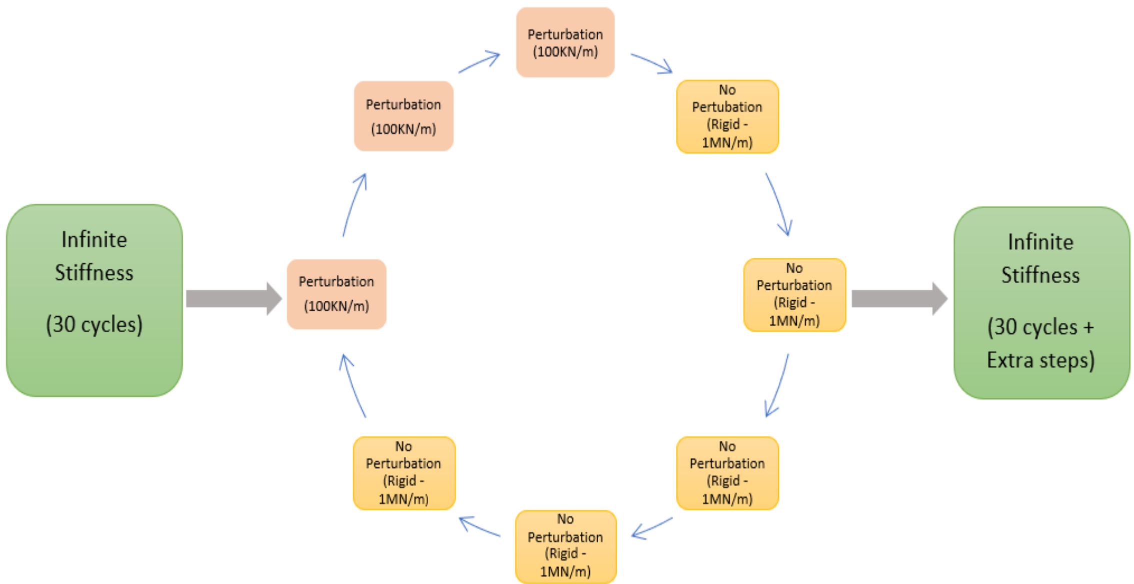

Figure 3.7: Experimental Protocol

walking such as pavement. After this, the cycle of alternate surface walking begins. There are 40 trials per compliant surface with 8 steps per trial, giving a total of 320 cycles. This is the period where the subject experiences the deflections on the left belt. Because enough data is needed to make good conclusions, a total of 40 trials were carried out per surface type. Perturbations were delivered following immediately after the left heel strike, approximately 5% of each gait cycle. Each perturbation delivered lasted throughout the left leg stance phase i.e. until the left leg toe is off. Every 5 steps on a solid surface (infinite stiffness) is followed by 3 steps on the compliant surface, a stiffness of 100kN/m for the first set and 60K/Nm for the second set. After 320 trials of alternating surface stiffness, the subject is then made to walk for a minimum of 30 cycles on a rigid surface, infinite stiffness. All subjects were made to wear a body harness as a safety measure, but no body weight support was used during the experiment.

3.4 Data Collection and Processing

This section discusses how data was collected during the experiment. The type of data necessary was; 1. Lower limb kinematics and 2. Muscle activation. Kinematic data was collected using the IR motion capture system whiles muscle activation was collected using wireless surface EMG electrodes.

3.4.1 Kinematic Data

Kinematic data collection for this experiment was needed not only to know the subjects joint angles but also for timing the changes in stiffness of the VST, as a way of determining when exactly a perturbation should occur within the subjects gait cycle. This allows for consistency across subjects.

Using an infrared (IR) motion capture system, kinematic data for both legs were collected at 140Hz. This system consists of two IR cameras (Code Laboratories Inc, model: DUO MINI LX) which track a total of 12 IR light emitting diode (LED) markers (Super Bright LEDs Inc, model: 1WS-850), six per each leg. The IR markers were placed as a pair each on the foot, the shank and the thigh at the lateral end of the leg, parallel to the sagittal plane. These markers provide information on the ankle, knee, and hip joint angles. The advantage of using the infrared-based measurement system is its ability to capture data in real time, as needed for this experiment. The cameras were carefully placed one on either side of the VST such that one captures the joint angles of the left leg while the other captures that of the right leg. This system provides automatic marker detection with high spatial and temporal resolution.

Prior to starting the experiment, the cameras are first calibrated to the length and position of the VST within which the subject would walk. For this experiment, the gait cycle begins with the left leg heel strike and continues until the next left leg heel strike i.e. when the tracked foot position reached its minimum value after full extension during the swing phase for both legs. Kinematic data was sampled at 140Hz and resampled at 0.1% of the gait cycle. This was then used to determine the timing of the gait cycles to process EMG data and for further processing of EMG data. The gait cycles were determined by finding the minimum of every foot position, closest to heel strike with a cutoff at 50.

3.4.2 Electromyographic Data

Muscle activity of the lower limb muscles, specifically the Tibialis Anterior (TA), Gastrocnemius (GA), Soleus (SOL) and Peroneus Longus (PL) were collected using a wireless surface EMG system (Delsys Trigno Wireless EMG) and recorded at 2000Hz. In this thesis, the focus is on the muscles that contribute to ankle dorsiflexion and plantarflexion as the ankle joint plays a very important role in stability. The placement of the surface EMG electrodes is shown in Figure 3.8.

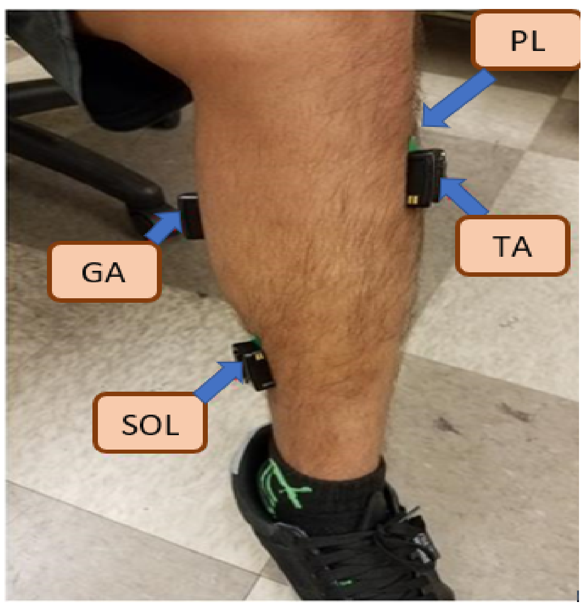

Figure 3.8: Surface EMG Sensor Placement for Lower Limb Muscles

The electrodes were placed on these muscles for both right and left legs of subjects.

Electrodes were correctly placed on these muscles and the clinical tests were done using seniam guidelines for EMG electrode placement[23]. The subjects performed the following; dorsiflexion of the ankle joint and inversion of the foot for the TA, heel up for the GA and SOL and plantarflexion of the ankle joint and eversion of the foot for the PL. Raw EMG signals were recorded at 2000Hz and using Simpsons 1/3 rule, the sum of the electric potential difference within each muscle was determined by finding the Root Mean Square(RMS) within a 250ms window for each muscle. Because of the need to compare the EMG amplitude of the different muscles for the different instances of surface stiffness perturbations, the signals were normalized to the maximum amplitude value for all data obtained for a specific muscle under the two experimental sets. Subsequently, the filtered EMG data was re-sampled at 0.1% of the gait cycle because of the dependence of muscle activity on the phase of the gait cycle. As the EMG data contained information for two instances of perturbed (walking on compliant surface) and unperturbed (walking on rigid surface of infinite stiffness), the data was further broken and separated into cycles of perturbed and unperturbed and was processed and plotted for two gait cycles, with the goal of investigating anticipatory responses. To determine statistically significant difference between rigid surface walking(unperturbed) and compliant surface walking(perturbed), two-tailed, two-sample unpaired t-tests were performed at 95% confidence level.

Figure 3.9 justifies the method employed in data segmentation for two gait cycles. It depicts the vertical displacement of the treadmill belt. The response of subjects is analyzed with respect to vertical deflection due to loading and lowered stiffness of the belt. The belt is made to deflect just before heel strike of the gait cycle. With the deflection of the belt, only downward motion is of importance as upward motion is due to oscillatory behavior. The mean and standard deviation of the unperturbed (rigid surface) is depicted in the red-dashed line while that of the perturbed transitions is

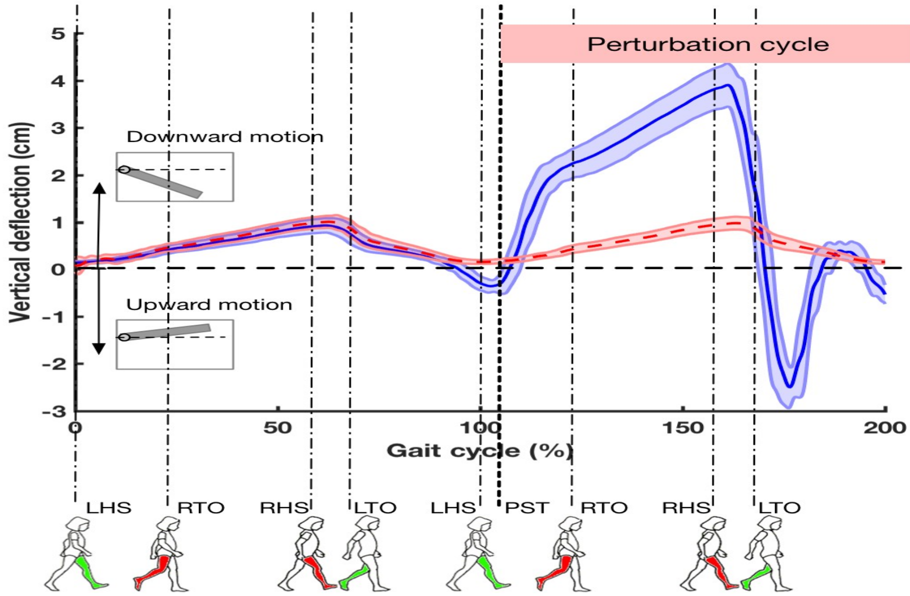

Figure 3.9: Vertical Deflection of the Left Belt of Treadmill across two Gait Cycles

indicated with a blue solid line. The perturbation occurs just after left leg heel strike at approximately 105% of the gait cycle.

Chapter 4

RESULTS AND DISCUSSION

This section presents and discusses the EMG activation plots obtained for the four lower limb muscles: Tibialis Anterior (TA), Gastrocnemius (GA), Soleus (SOL) and Peroneus Longus (PL) muscles for both limbs. Data was processed and analyzed for four out of a total of 6 subjects that participated due to some discrepancies in results and other factors that could potentially affect validity. Plots indicated in this section are of a representative subject. In all plots from subsection 4.1 -4.4, the solid blue line represents the response to perturbed cycles, the solid red line represents the response to unperturbed cycles and the statistically significant difference is indicated by solid magenta lines at the top of the graph. They indicate muscle response for both legs the perturbed leg(left) and the unperturbed leg (Right). In this section, four main questions are addressed;

1.) Is there anticipatory behavior in EMG while transitioning from a rigid to a compliant surface?

2.) Is there reactive behavior in EMG while transitioning from a rigid to a compliant surface?

3.) Is the result in (1) affected or influenced by the surface type?

4.) Is the result in (2) affected or influenced by the surface type?

Subsections 4.1 and 4.2 discusses the activity of the perturbed leg, that which experiences surface perturbations for 60KN/m and 100KN/m respectively in response to questions (1) and (2). Subsections 4.3 and 4.4 also expands on the activity of the unperturbed leg, that which does not experience changes in surface stiffness for 60KN/m and 100KN/m respectively in response to questions (1) and (2). Finally, subsection 4.5 discusses the activity of both perturbed and unperturbed legs in response to question (3) and (4). During the experiments, the VST was commanded to deliver a stiffness of 60KN/m which is the least compliant in comparison to a stiffness of 100KN/m. The first gait cycle (0-100%) represents the step before perturbation occurs and the second gait cycle (100% to 200%) represents the step after the perturbation occurs. As stated earlier, the perturbation occurs at approximately 105% of the gait cycle which is indicated by PST in the plots. All subjects were verbally informed prior to a perturbation to best simulate real-world experience. To conclude on anticipatory behavior, statistically significant difference at 95% confidence level must be observed prior to a perturbation i.e. before 105% of the gait cycle. A reactive response is considered for statistically significant difference after the perturbation.

4.1 Muscle Responses to Compliant Surface 1 (60KN/m Stiffness) for the Left (Perturbed) Leg

Statistically significant difference is observed in all four muscles. The TA and PL indicate anticipatory response at about 98% and 102% in Figure 4.1 and 4.4 respectively. There is no significant anticipatory response in the GA and SOL muscles. All four muscles present a reactive response just after the perturbation occurs. For the plantar flexors of the ankle joint, the GA, SOL and PL, there is an increase in activation of the muscle in response to the compliant surface. This is noticed right after the perturbation occurs. This indicates the role of the TA and PL during transitioning as the subject prepares to step on a compliant surface. Activation is increased in the plantar flexors to maintain balance over the compliant surface. Additionally, the significant rise in the TA perturbed response towards the very end of the second gait cycle of the left leg shows the possibility of its behavior just before stepping on the compliant surface with the assumption of three gait cycles plotted where the third also indicates perturbation.

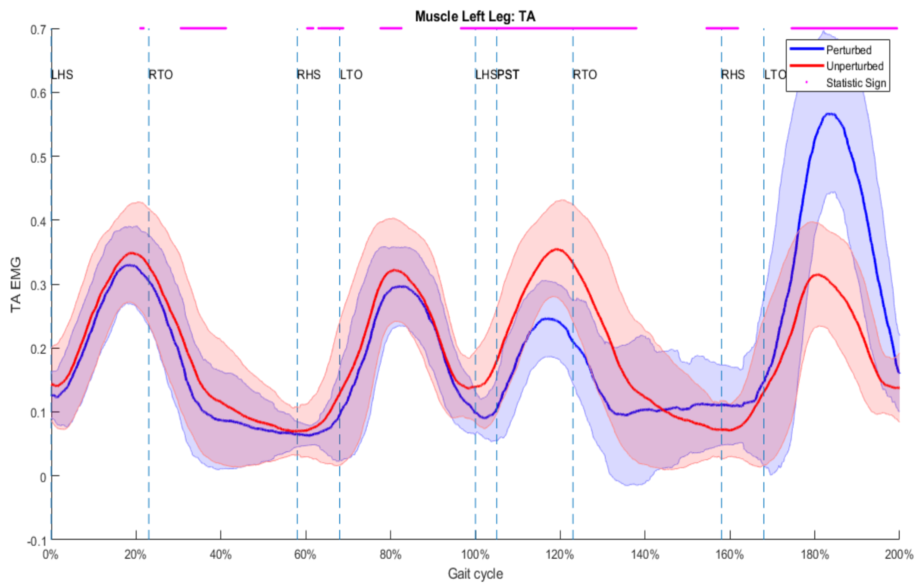

Figure 4.1: Response of Left Tibialis Anterior (TA) to 60KN/m Perturbation

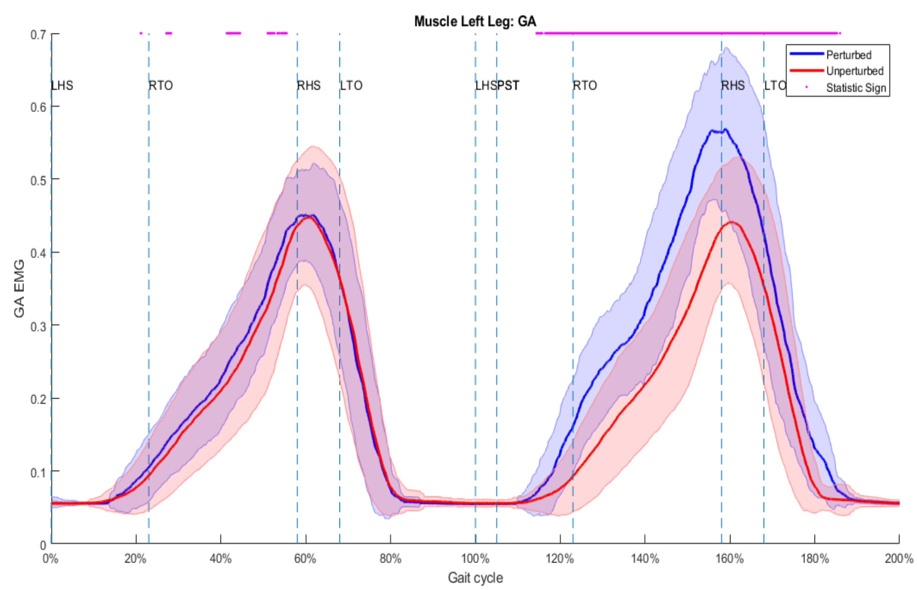

Figure 4.2: Response of Left Gastrocnemius (GA) to 60KN/m Perturbation

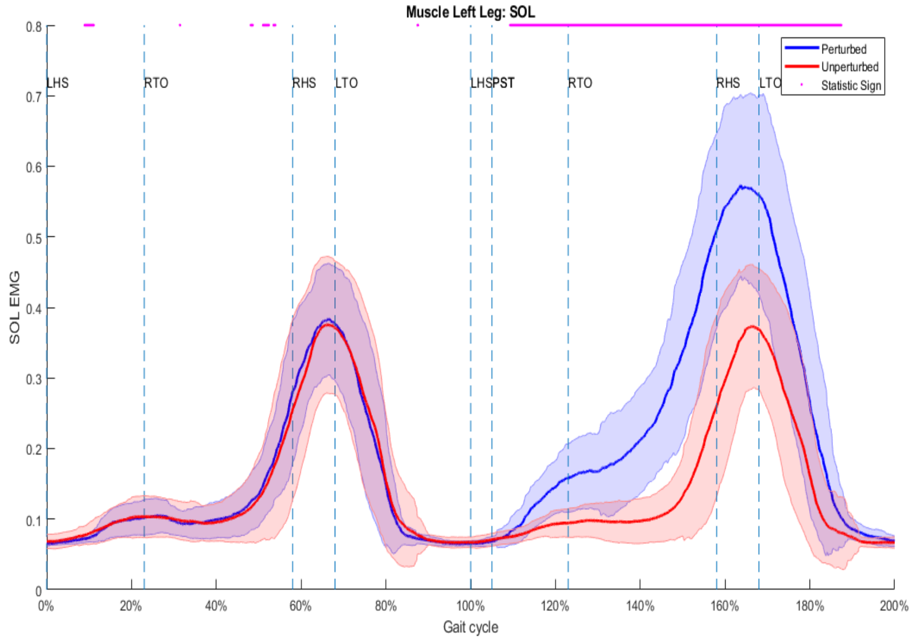

Figure 4.3: Response of Left Soleus (SOL) to 60KN/m Perturbation

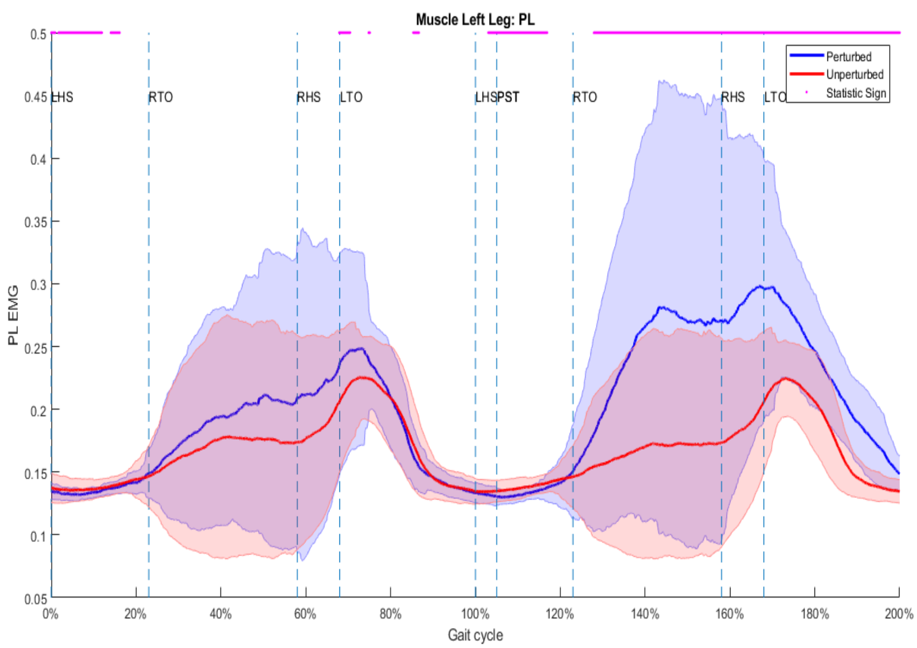

Figure 4.4: Response of Left Peroneus Longus (PL) to 60KN/m Perturbation

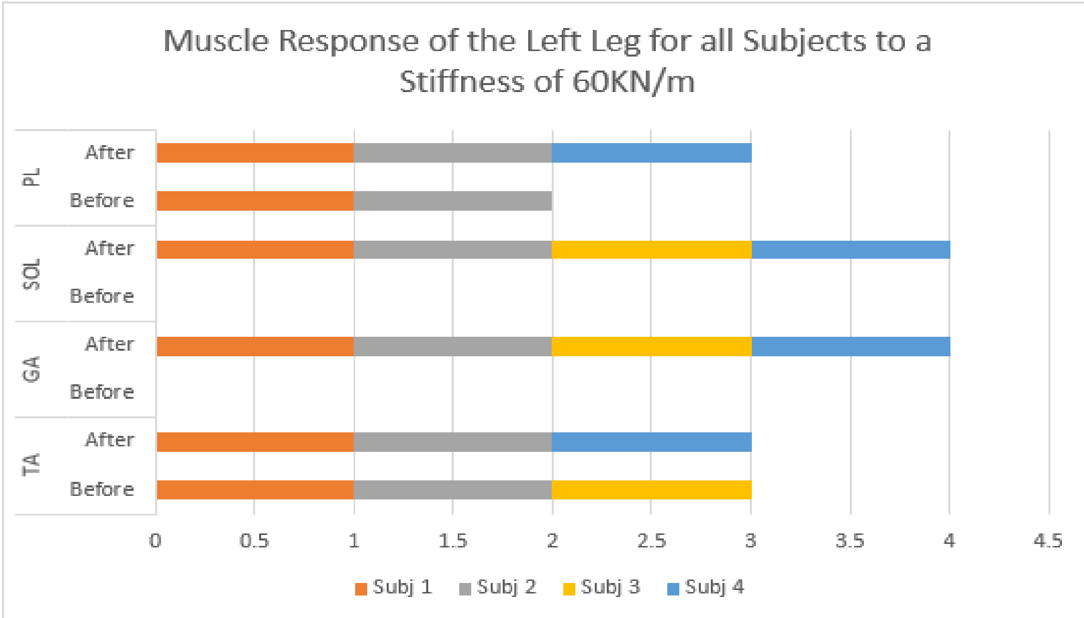

Figure 4.5: Left Leg Anticipatory and Reactive Response for all Four Subjects to 60KN/m Stiffness

Figure 4.5 depicts the response distribution for all four subjects. Anticipatory response is observed in the TA and PL for most subjects and reactive response is seen in all four muscles.

4.2 Muscle Responses to Compliant Surface 2 (100KN/m Stiffness) for the Left Leg

In this section as well, anticipatory and reactive responses are observed for the left leg to a perturbation of 100KN/m. Anticipatory response is observed in the TA and PL and reactive response is observed for all four muscles as depicted in Figures 4.6 to 4.9. Like in the response to a stiffness of 60KN/m discussed in subsection 4.1, the patterns are very similar. It can then be said that the brain prepares itself to tackle compliant surfaces in a rather similar manner. In these figures as well, the GA, TA and SOL show a rise in perturbed muscle activation after the perturbation and this is expected because of their role in plantar-flexion of the ankle during push-off.

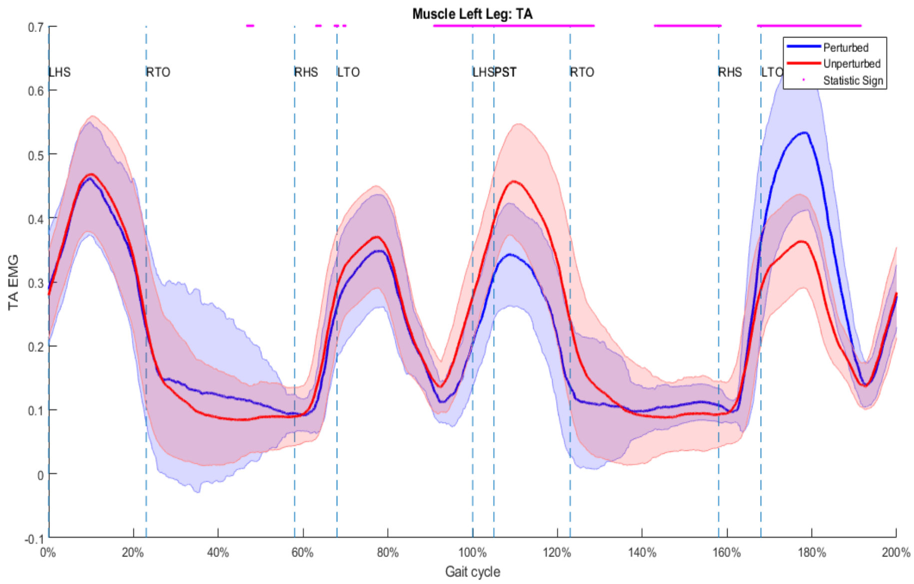

Figure 4.6: Response of Left Tibialis Anterior (TA) to 100KN/m Perturbation

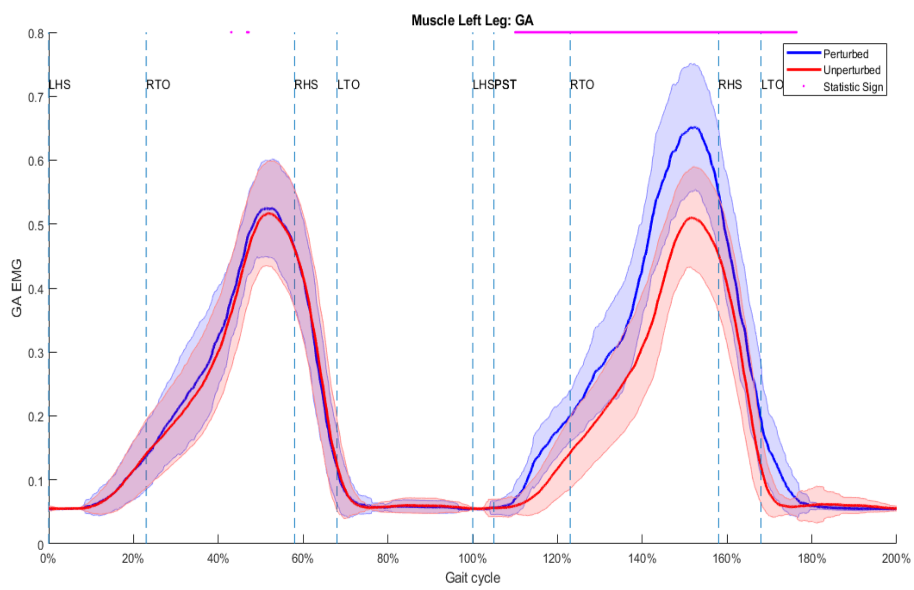

Figure 4.7: Response of Left Gastrocnemius (GA) to 100KN/m Perturbation

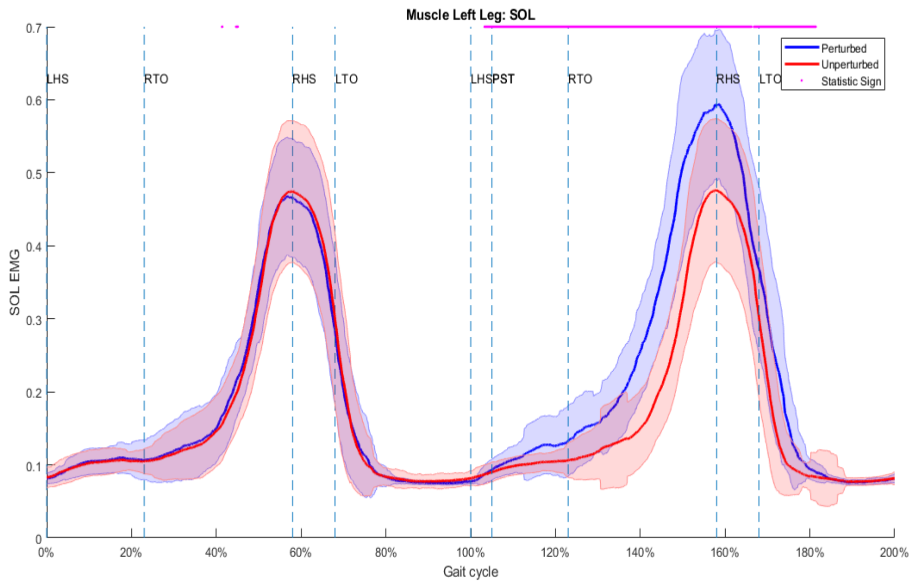

Figure 4.8: Response of Left Soleus (SOL) to 100KN/m Perturbation

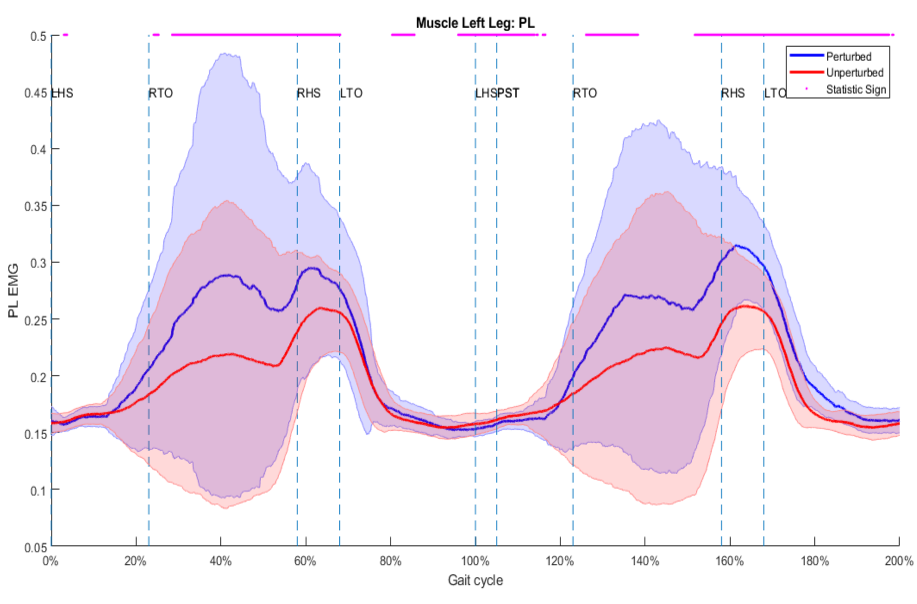

Figure 4.9: Response of Left Peroneus Longus (PL) to 100KN/m Perturbation

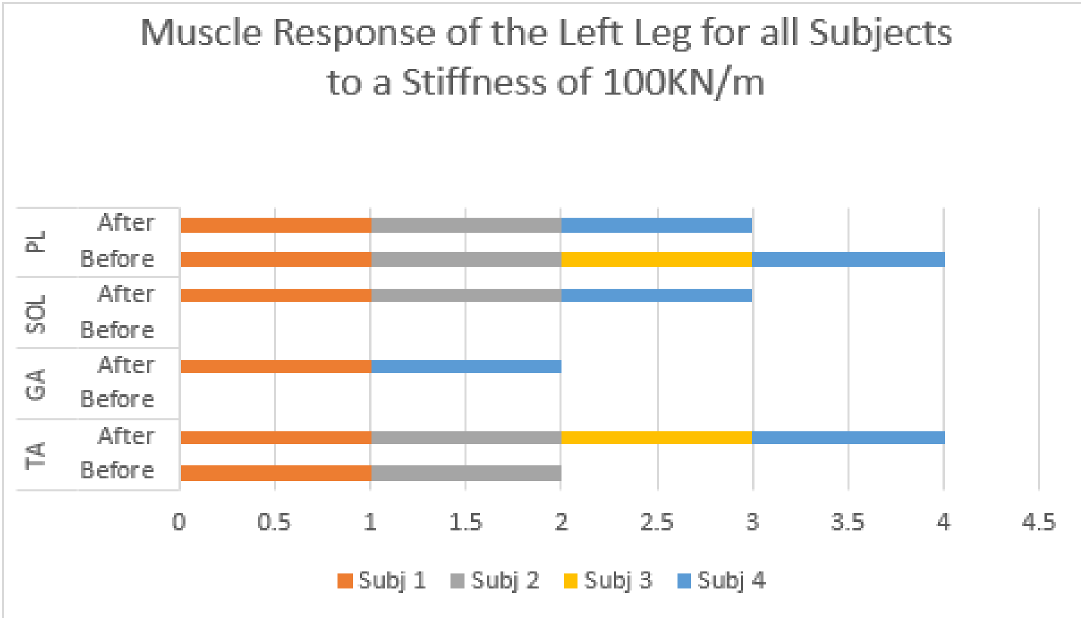

Figure 4.10: Left Leg Anticipatory and Reactive Response for all Four Subjects to 100KN/m Stiffness

4.3 Muscle Responses to Compliant Surface 1 (60KN/m Stiffness) for the Right Leg

This section discusses the response of the right leg to perturbation which occurs on the left leg. While this leg, does not experience change in surface stiffness, it is useful to understand how a change in surface stiffness of the other leg affects muscle activation. It is expected that, a deflection in the left belt causes some variability in the gait of the right leg. Of notable significance is the strong anticipatory and reactive behavior in the soleus (SOL) muscle just before and after the perturbation and the anticipatory behavior present in the gastrocnemius (GA) in Figure 4.12 and 4.13 respectively. However, it can be observed that the increase in activation during perturbed cycles of the left leg is not that high but for the case of the GA.

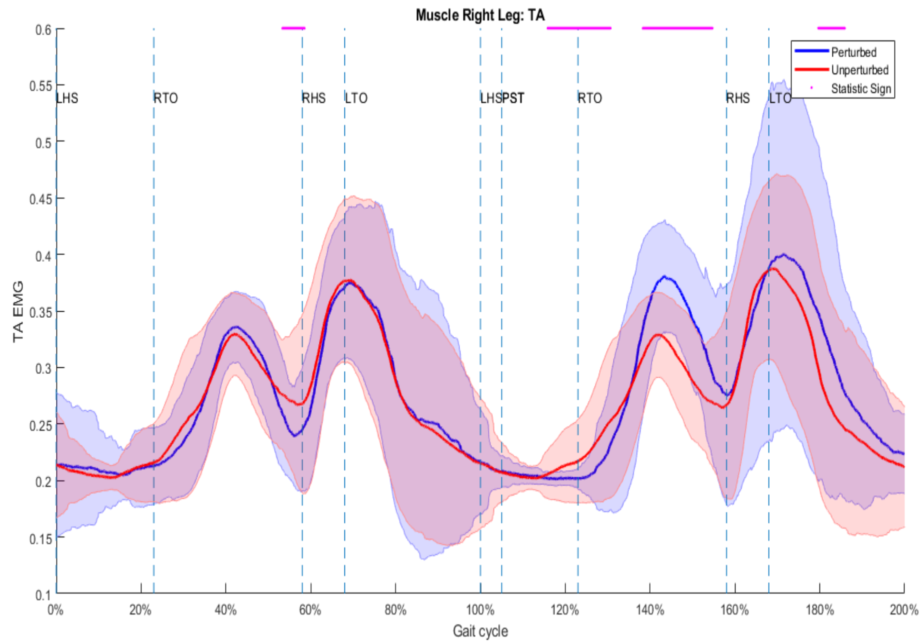

Figure 4.11: Response of Right Tibialis Anterior (TA) to 60KN/m Perturbation

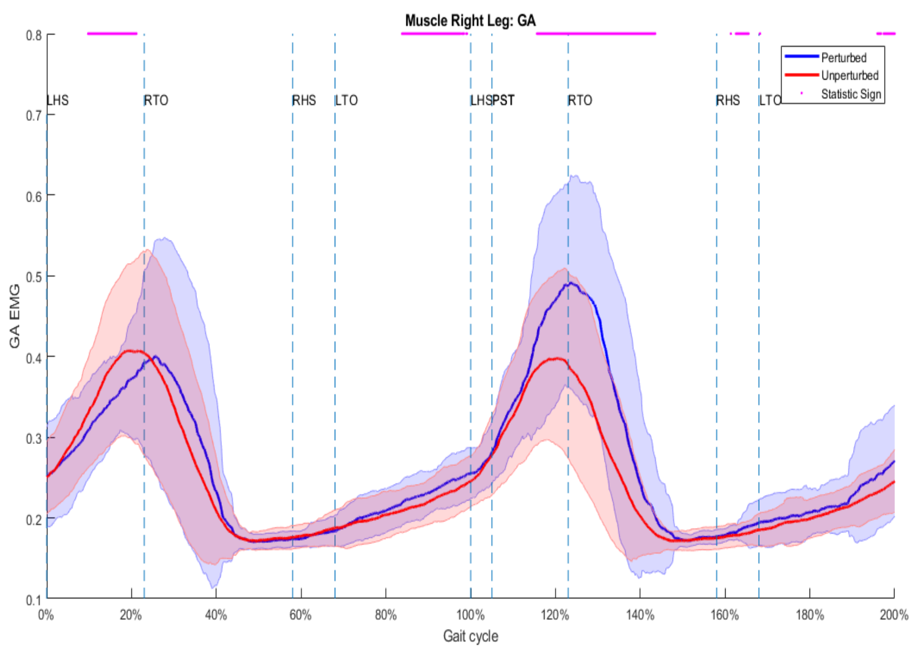

Figure 4.12: Response of Right Gastrocnemius (GA) to 60KN/m Perturbation

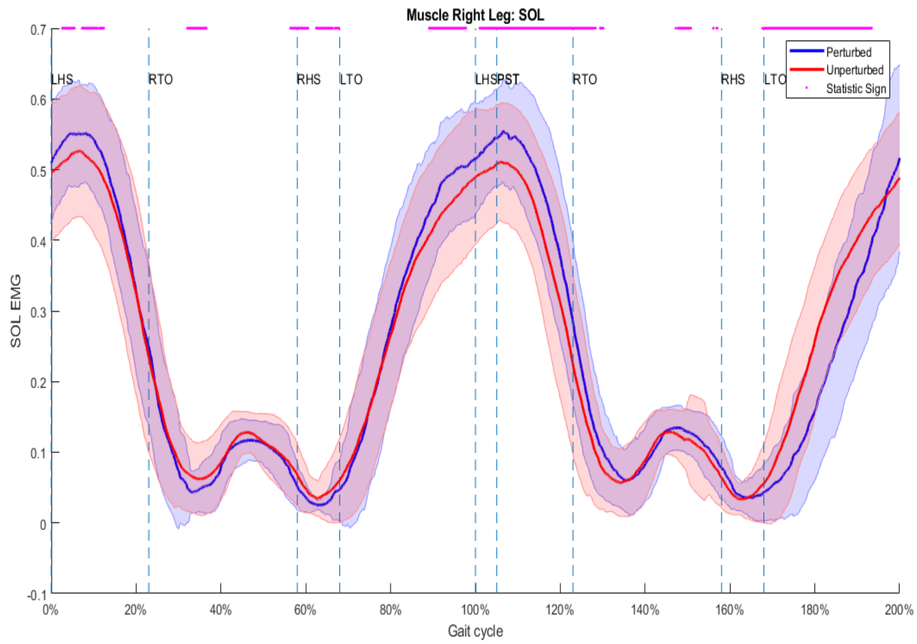

Figure 4.13: Response of Right Soleus (SOL) to 60KN/m Perturbation

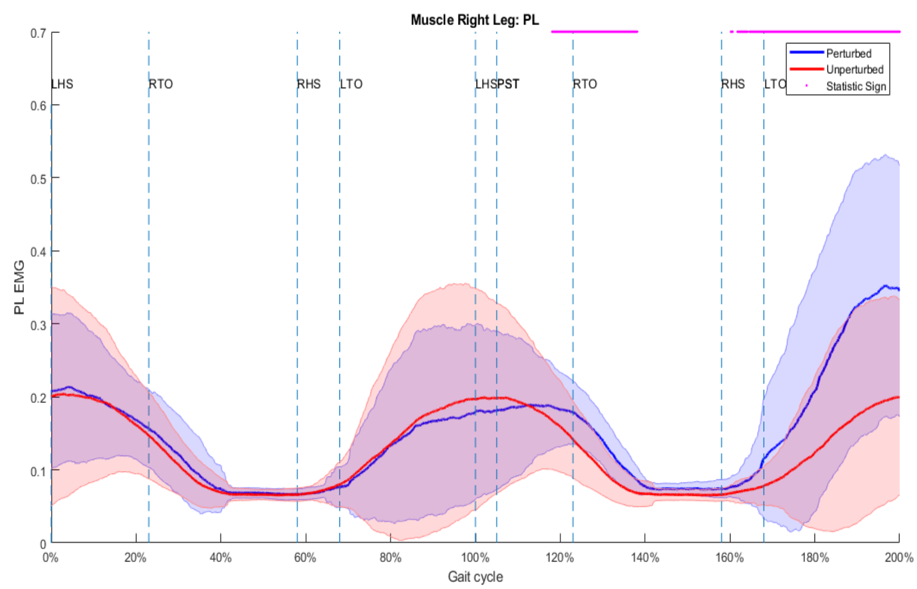

Figure 4.14: Response of Right Peroneus Longus (PL) to 60KN/m Perturbation

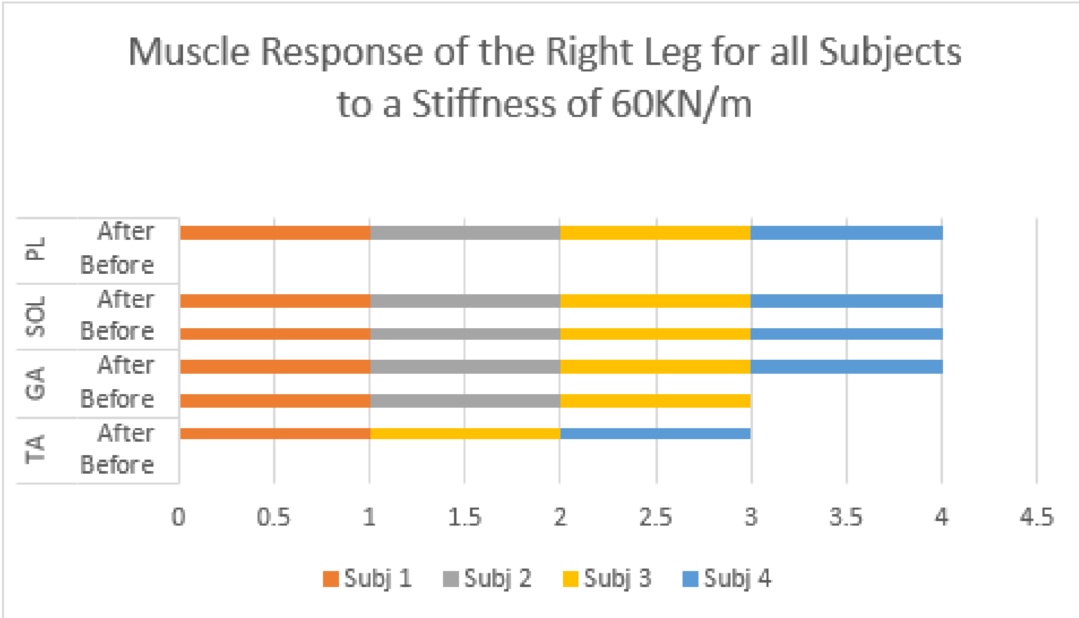

Figure 4.15: Right Leg Anticipatory and Reactive Response for all Four Subjects to 60KN/m Stiffness

4.4 Muscle Responses to Compliant Surface 2 (100KN/m Stiffness) for the Right Leg

Statistically significant difference is observed in the TA and SOL before perturbation occurs in the left leg in Figures 4.16 and 4.18 respectively. From sections 4.3 and 4.4, we see that information from the right leg can potentially be used to encode information regarding anticipatory strategies prior to stepping on compliant surfaces.

Notably in section 4.3 and 4.4, from the plots, we can deduce that there is not much variation in amplitude considering the two modes, there still is a difference and further investigation into this can be used to fine tune movements of the prosthetic limb in the case of bilateral lower limb amputations.

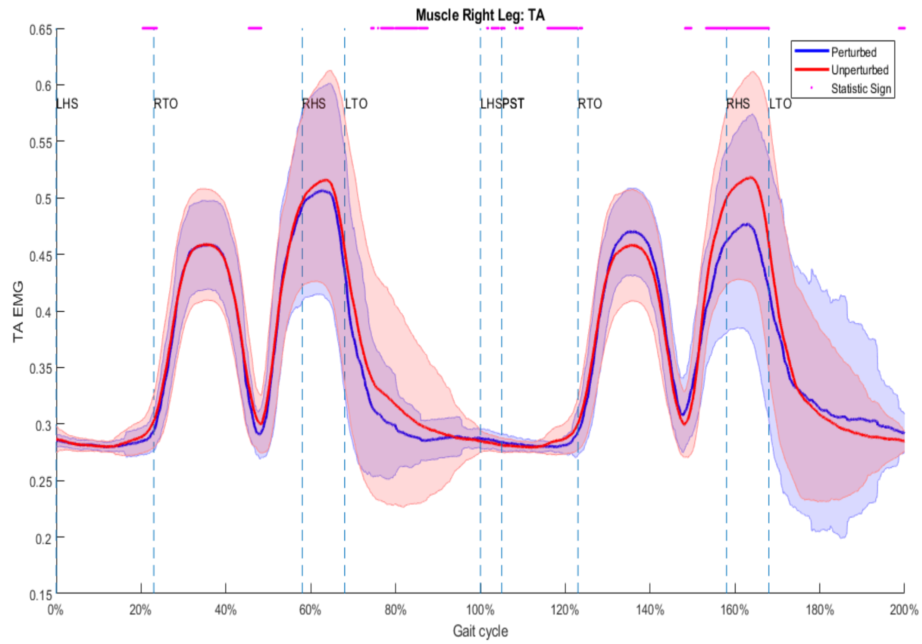

Figure 4.16: Response of Right Tibialis Anterior (TA) to 100KN/m Perturbation

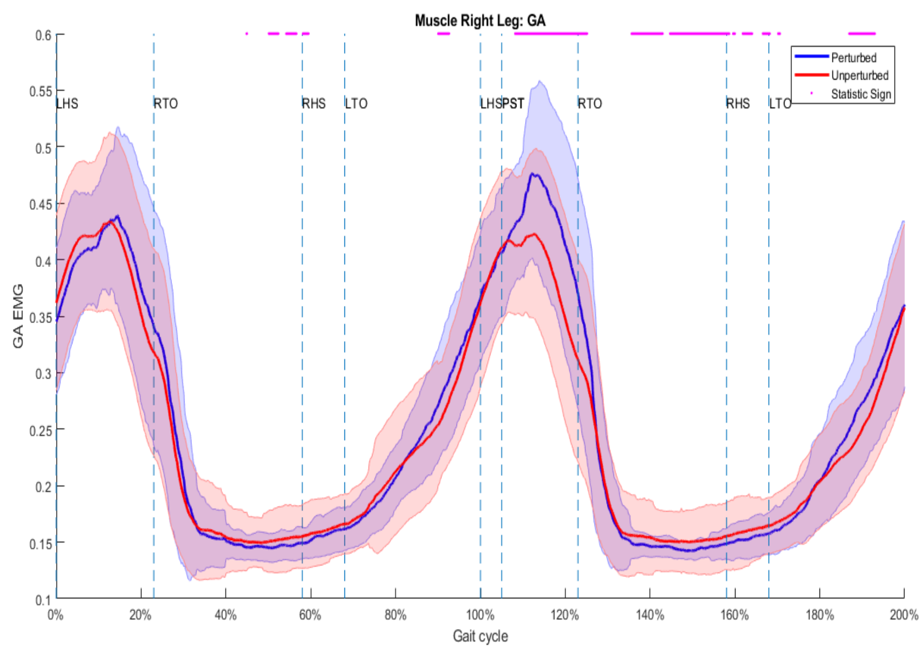

Figure 4.17: Response of Right Gastrocnemius (GA) to 100KN/m Perturbation

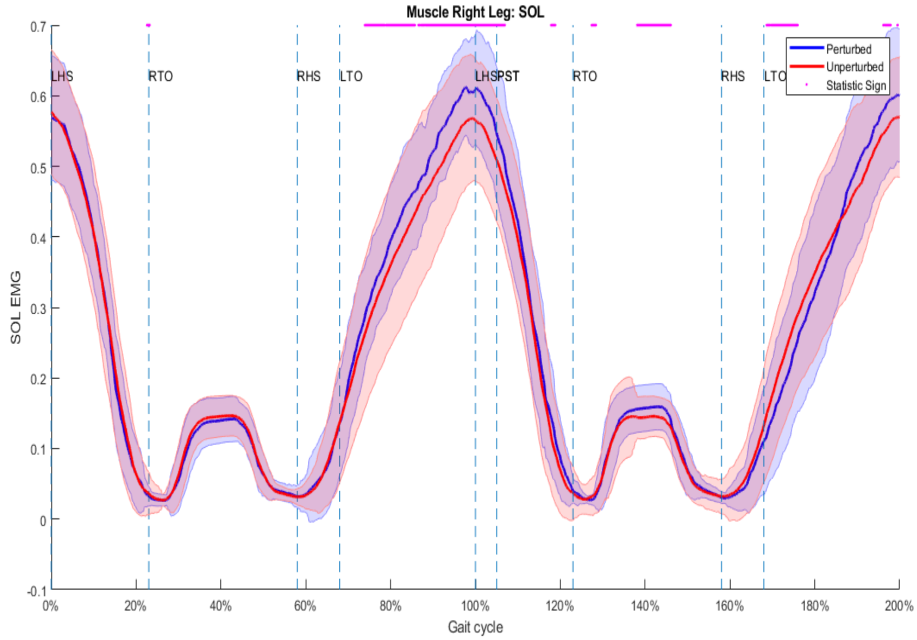

Figure 4.18: Response of Right Soleus (SOL) to 100KN/m Perturbation

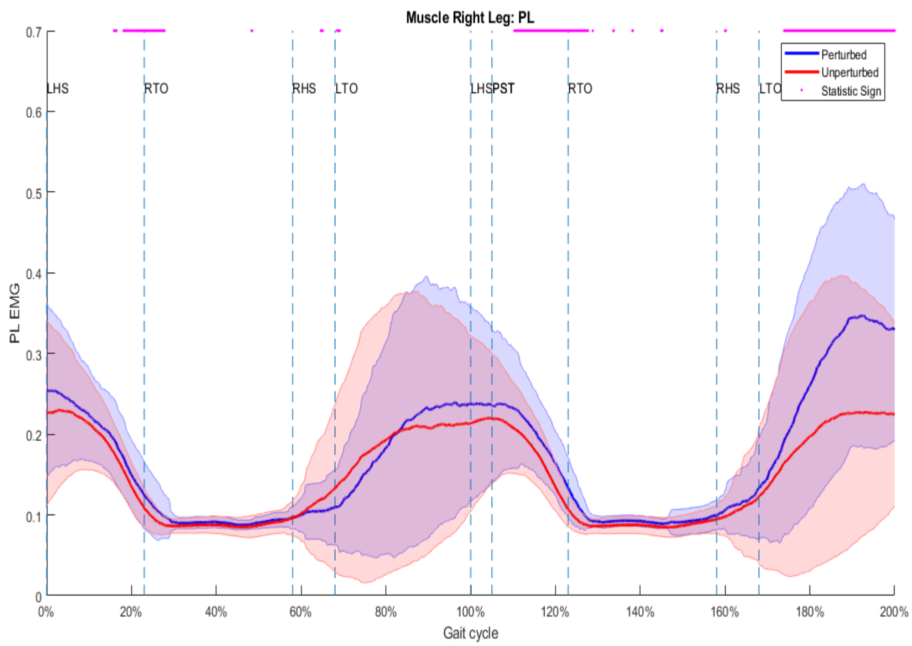

Figure 4.19: Response of Right Peroneus Longus (PL) to 100KN/m Perturbation

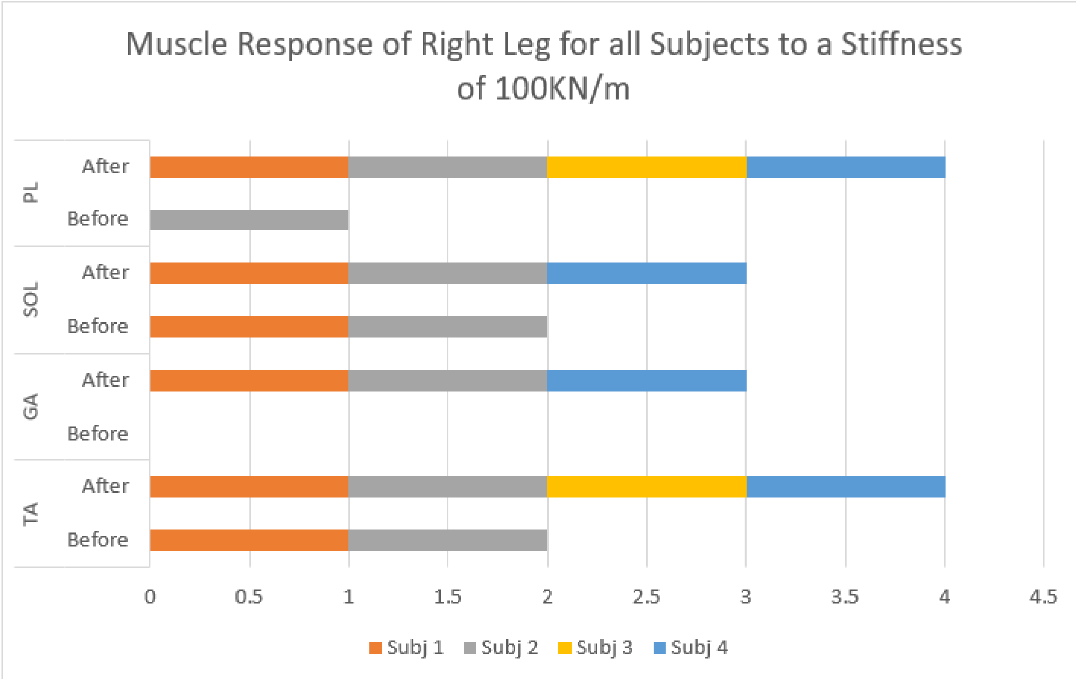

Figure 4.20: Right Leg Anticipatory and Reactive Response for all Four Subjects to 100KN/m Stiffness

4.5 Comparison of Anticipatory and Reactive Response for the Left (Perturbed) Leg to both Surfaces (60KN/m and 100KN/m)

In this section, the response of each muscle to the different surface modes are plotted. For each graph, the solid blue line and the solid red line represent perturbed and unperturbed cycle responses to a stiffness of 100KN/m (the least compliant surface) respectively. The solid cyan line and the solid yellow line represent perturbed and unperturbed cycle responses to a stiffness of 60KN/m (the most compliant surface) respectively. In essence, the dark-colored shaded graph represents the response to stiffness of 100KN/m and the light-colored shaded graph represents the response to a stiffness of 60KN/m. In all instances, it can be clearly seen that muscle response varies for the two surfaces which is expected as muscles will not be activated at the same level for different surfaces although the pattern must be the similar. In figure

4.21 we observe the response of the ankle dorsiflexor, the Tibialis Anterior (TA) to the two surfaces. Anticipatory and reactive behavior exists prior to and after perturbation occurs at 105% of the gait cycle, lasting from approximately 95% to 140% of the gait cycle.The TA tends to indicate a much stronger activation for the more compliant surface at about 190% of the gait cycle where we see a higher activation on the more compliant surface. Additionally, we see in the GA, SOL and PL that in after stepping on a compliant surface, there is an increase in muscle activation. The ankle plantar flexors, therefore, increase activation on compliant surfaces. However, the more compliant the surface is, the less activation we have in the plantar flexors.

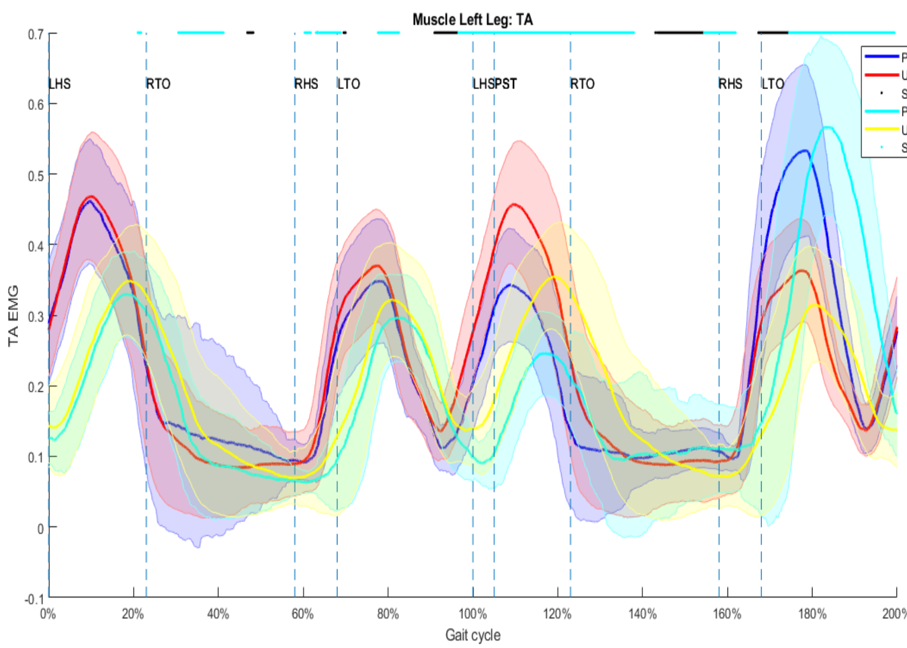

Figure 4.21: Left Tibialis Anterior (TA) Response to both Compliant Surfaces

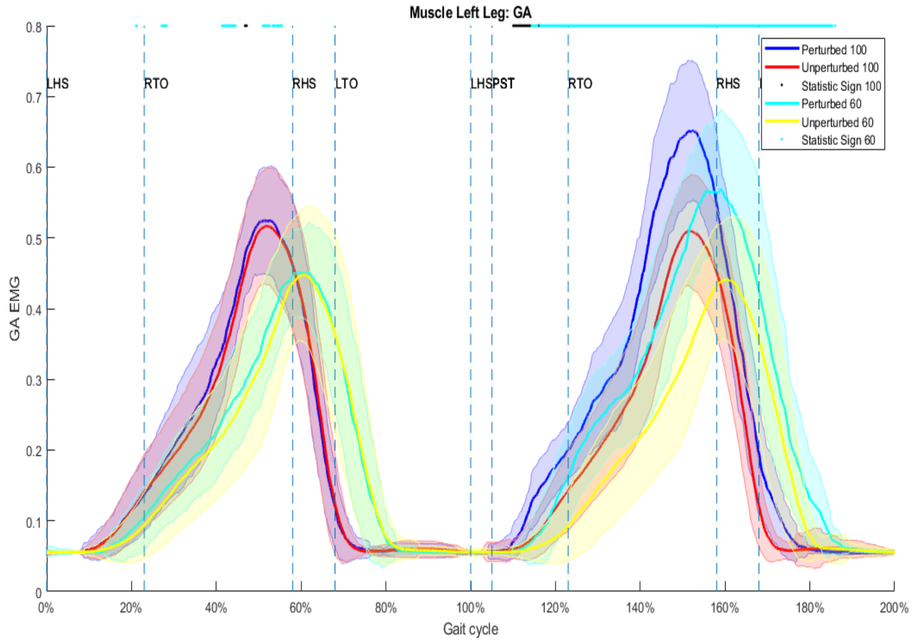

Figure 4.22: Left Gastrocnemius Response (GA) Response to both Compliant Surfaces

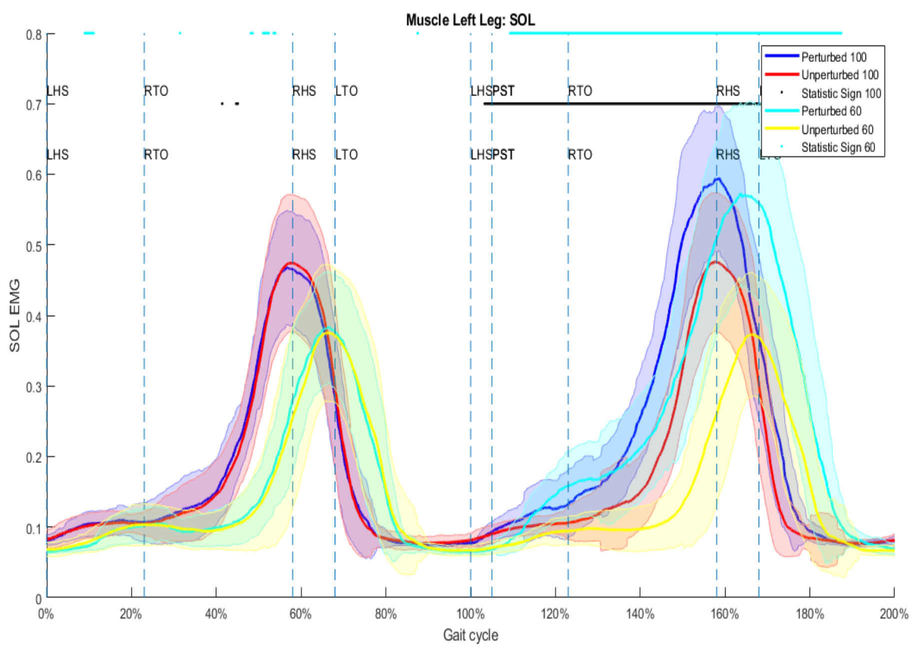

Figure 4.23: Left Soleus (SOL) Response to both Compliant Surfaces

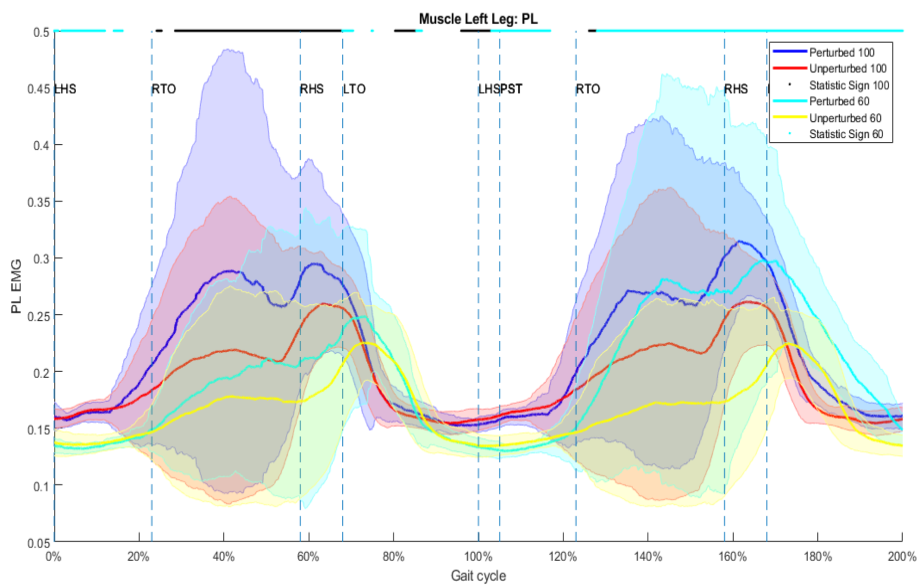

Figure 4.24: Left Peroneus Longus (PL) Response to both Compliant Surfaces

Chapter 5

CONCLUSION AND FUTURE WORK

5.1 Conclusion

The primary goal of this work was to identify anticipatory behaviour of specific lower limb muscles - tibialis anterior(TA), gastrocnemius(GA), soleus(SOL) and peroneus longus(PL) muscles while humans transition during walking from a rigid surface to a compliant surface for example from pavement to beach sand and to determine whether the response varied based on the surface type. By investigating two different compliant surface levels - most compliant(60KN/m stiffness) and least compliant(100KN/m), it can be inferred based on results obtained, that indeed there is anticipatory response and the response varies based on the surface type as intact humans prepare to make a step on a compliant surface. Most specifically, there was a decrease in activation of the TA right before the perturbation and an increase during the terminal swing. In comparing the response of the TA for the two compliant surface types, there is a decrease in activation the more compliant the surface is. However, there is a higher increase in activation of the most compliant surface above what is observed in the least compliant surface. On the other end, the GA, SOL, and PL show an increase in activation after stepping on a compliant surface with GA showing the highest level of muscle activation. As these muscles are all ankle plantar flexors, their response is quite similar. From the viewpoint of the two compliant surfaces, the GA, PL, and SOL showed higher activation in response to the least compliant surface. Hence, the more complaint the surface is, the less the muscle is activated. Statistically significant difference prior to stepping on a compliant surface is observed in the TA and PL while the reactive response is present in all four muscles.

5.2 Recommendation for Future Work

In this study, anticipatory muscle responses to changes in surface compliance were investigated to be implemented in the control of powered ankle-foot prostheses. Understanding the response of intact subjects was necessary as the base level information. To further advance knowledge in this area to the point of its implementation, future studies will be to compare muscle responses between intact and non-intact subjects. As the end beneficiaries are amputees, an understanding of their response as it relates to intact subjects is of relevance. Most particularly, studies should be carried out with both right and left unilateral amputees as well as intact subjects. This will be useful in determining whether there is the for a separate kind of investigation for amputees, or whether the results obtained will be implemented in the control of prostheses for increased robustness.

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APPENDIX A

.1 APPROVAL DOCUMENTS FOR EXPERIMENT

APPROVAL:CONTINUATION

Panagiotis Artemiadis
Engineering of Matter, Transport and Energy, School for (SEMTE)
480/965-4182
Panagiotis.Artemiadis@asu.edu

Dear Panagiotis Artemiadis:

On 3/30/2018 the ASU IRB reviewed the following protocol:

Type of Review:	Modification and Continuing Review
Title:	Investigation of human gait using a variable stiffness treadmill
Investigator:	Panagiotis Artemiadis
IRB ID:	STUDY00001001
Category of review:	(4) Noninvasive procedures
Funding:	Name: National Science Foundation (NSF), Grant Office ID: FP10646, Funding Source ID: Proposal CMMI-1727838
Grant Title:	None
Grant ID:	None
Documents Reviewed:	• Recruitment Script Verbal.pdf, Category: Recruitment Materials; • Informed Consent.doc, Category: IRB Protocol; • citiCompletionReport2612632_Artemiadis.pdf, Category: Other (to reflect anything not captured above); • HRP-503b-%20TEMPLATE%20PROTOCOLBioscience_v4.docx, Category: IRB Protocol; • Recruitment Flyer.pdf, Category: Recruitment Materials; • Informed Consent.pdf, Category: Consent Form; • Recruitment Script Email.pdf, Category: Recruitment Materials;

Page 1 of 2

The IRB approved the protocol from 3/30/2018 to 4/23/2019 inclusive. Three weeks before 4/23/2019 you are to submit a completed Continuing Review application and required attachments to request continuing approval or closure.

If continuing review approval is not granted before the expiration date of 4/23/2019 approval of this protocol expires on that date. When consent is appropriate, you must use final, watermarked versions available under the "Documents" tab in ERA-IRB.

In conducting this protocol you are required to follow the requirements listed in the INVESTIGATOR MANUAL (HRP-103).

Sincerely,

IRB Administrator

cc:
Jeffrey Skidmore
Ryan Frost
Danielle Stevens

APPENDIX B

.2 COPYRIGHT PERMISSION

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Author:	Jeffrey Skidmore
Publication:	Mechatronics, IEEE/ASME Transactions on
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