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Neural and Mechanical Changes for Adapting Joint Mechanics in Different Environments

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ABSTRACT

Neural and Mechanical Changes for Adapting Joint Mechanics in Different Environments

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Humans have a remarkable ability to walk on a variety of surfaces. Compliant, uneven, or even slippery surfaces present little challenge to most people, yet are hazardous to individuals with locomotor impairments and even to bipedal robotic systems designed to mimic what we understand about human locomotion. Our ability to navigate seamlessly across different terrains stems in part from how we can adapt the mechanical properties of our legs to the unique requirements of each surface. The objective of this dissertation was to study this ability, using locomotion on slippery surfaces as a paradigm for examining the neural and mechanical adaptations that allow us to traverse a multitude of terrains.

We demonstrated a significant adaptation for walking on slippery surfaces is to reduce ankle muscle activity, which directly contributes to a reduction in shear forces and ankle joint stiffness, minimizing slip potential. We further investigated how individual changes in joint torque and joint position, which change simultaneously during walking, affect ankle joint mechanics to gain a better understanding of the link between neural and mechanical adaptations during walking. We found that isolated changes in joint position and joint torque reduced ankle joint stiffness, and simultaneous changes resulted in a dependence on the direction of these changes. Our work demonstrates the neural and mechanical adaptations employed on unique terrains, such as on a slippery surface, are critical for successfully negotiating the terrain and reducing the likelihood of a fall. This has significant implications for people who have impaired neural control and for whom the mechanical properties of the leg have been altered through injury or disease.

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Dedication

To Chris, for your unwavering support and perfect calves.

TABLE OF CONTENTS

List of Tables

LIST OF FIGURES

1. INTRODUCTION

1.1. Motivation

In our everyday lives, we are continuously confronted with variable surface conditions. Whether you step off a concrete sidewalk onto a patch of grass or go for a long stroll on a sandy beach, we must continuously adapt our gait for different terrains. The ultimate consequence for failing to adapt appropriately to a new terrain is the initiation of a fall. Falling is a critical problem, responsible for the largest number of injury-related emergency room visits [1] and the second largest source of death caused by accidental injury [2] in the United States. Thus, the economic cost of falling is substantial, with direct costs of $8.6 billion in 2011 [3]. Some people experience a higher prevalence of falls, including older adults [4] and individuals with a lower-limb amputation [5]. Fear of falling is substantial for both populations, contributing to activity avoidance and a reduced quality of life [5, 6]. Understanding how we appropriately adapt our gait and avoid falls across different terrains is important for providing insight into why certain people have a higher prevalence of falls and what can be done to improve their interactions with variable walking conditions.

Our ability to adapt to new terrains is, in part, facilitated by modulating our limb impedance. Joint impedance defines the dynamic relationship between an imposed change in joint angle and the torques generated in response, which essentially describes how our body reacts to unexpected disturbances in our environment. When walking across variable terrains, we continuously adapt our lower-limb kinematics (joint angles) and kinetics (joint torques). Ultimately driving these adaptations is the neural control, which coordinates which joint to move at what time and by how much. We can use surface electromyography (EMG) to record the activity of individual muscles to gain insight into the neural control of these variables during movement. Under isometric (constant joint angle) and isotonic (constant joint torque) conditions, a strong dependence between muscle activation and joint stiffness, the static component of joint impedance, has been demonstrated [7, 8]. This relationship has been useful for inferring the mechanical consequences of changes in muscle activity when direct measurements of impedance cannot be made.

Walking on a slippery surface is one terrain in which observed changes in muscle activity have been implicated as having a mechanical consequence. Appropriately adapting our gait in the presence of a slippery surface is critical as the consequences of inappropriate adaptations are dire, including severe injury and death. Up to half of all fall-related injuries are the result of a slip [9], so understanding how slip-related falls can be avoided will contribute significantly to an overall reduction in falls.

While the majority of previous slipping studies have assessed the factors that contribute to a successful recovery after a slip that avoids a fall [4, 10-13], no study has identified how people naturally adapt to the slippery surface to avoid the initiation of a slip, subverting any potential for injury. Chambers and Cham attempted to identify how people reduce slip potential by telling subjects they might encounter a single slippery spot and monitoring the changes that occurred when they walked across a normal, dry surface [14]. They found that when anticipating stepping onto a slippery spot, people increased cocontraction of the muscles spanning the ankle and knee, suggesting people were modulating their ankle and knee joint stiffness. They concluded that this strategy helped reduce slip potential, which was quantified as the reduction in the required coefficient of friction. Importantly, this strategy was not evaluated in the presence of a slippery surface, so it is unclear whether this reduction in slip potential would result in slip avoidance. Work by Cappellini and colleagues suggests increased muscle activity will not result in slip avoidance. They assessed the changes in lower-limb muscle activity that occurred during initial exposure to a slippery walkway and showed that lower-limb muscle activity again increased, yet subjects continuously slipped throughout the trials [15]. Cappellini and colleagues also suggested people were adopting a limb-stiffening strategy to negotiate the slippery surface, but it is unclear whether this strategy is appropriate as it sill resulted in slipping that could have triggered falls. While these studies did not evaluate slip avoidance strategies on slippery surfaces, they did highlight the potential importance of modulating lower-limb muscle activity and brought attention to the possible role of joint impedance in preventing slipping.

Though it is common to infer changes in joint mechanics based on changes in muscle activation [14-18], whether this inference is valid during movement has not been confirmed. The few studies that evaluated joint impedance during movement tasks would suggest that this inference is invalid [19-22]. Specifically, studies that evaluated impedance during voluntary movement consistently showed that the relationship between stiffness and muscle activation or joint torque was complex or non-existent and stiffness was substantially lower during movement than during posture. The complexity in the relationship observed between stiffness and muscle activation is likely because the cited studies were not designed to evaluate this relationship while controlling for the factors that are known to influence impedance, such as joint position and joint torque, as was done during the postural studies. Because these factors have not been controlled for, we still do not know how impedance is modulated by changes in muscle activation during movement.

The objective of this dissertation was to study our ability to navigate seamlessly across different terrains, using locomotion on slippery surfaces as a paradigm for examining the neural and mechanical adaptations that allow us to traverse a multitude of terrains. We chose to investigate the neural strategies and mechanical consequences of walking on a slippery surface because of the significant consequences associated with the use of inappropriate strategies and because it allowed us to investigate the mechanical consequences of a change in muscle activation for a natural task under relatively well-controlled conditions. We first aimed to fill the gap in previous slipping literature by evaluating the gait adaptations, in particular adaptations in lower-limb muscle activity, that are required for unimpaired individuals to walk continuously on slippery surfaces without slipping, which was addressed in Chapter 2. We then investigated the consequences of changes in ankle muscle activation observed in Chapter 2 by determining its effect on the shear forces and ankle joint stiffness, which was addressed in Chapter 3. Finally, the relationship between muscle activation and stiffness observed in Chapter 3 was different than that during posture and the reason for this is still poorly understood. In our final experiment, we further explored the differences between impedance modulation during posture and movement, by isolating the effect of a changing joint position and changing joint torque, which simultaneously change during walking, on stiffness. The findings from this study are addressed in Chapter 4.

The remainder of this introduction provides background information on what is currently known of how we interact with slippery surfaces and the work that remains to be done for understanding how we can improve the outcomes of these interactions. We discuss implications of modulating lower-limb muscle activity on slippery surfaces and the potential mechanical consequences. Finally, we discuss what is known of the modulation of joint impedance during movement and where these findings have digressed from the postural studies. We outline the need for a more well-controlled study so that we can understand how impedance is modulated during movement compared to posture.

1.2. Background

1.2.1. Interactions with Slippery Surfaces

There has been a substantial amount of research done on the biomechanics of walking on slippery surfaces [23]. The primary focus of early studies was to evaluate the kinematics and kinetics produced during recovery from an unexpected slip and to determine whether there were any factors that predisposed individuals to fall. In general, these studies identified the *reactive strategies* used to compensate after a slip. In contrast, other researchers have been interested in how prior awareness of or previous exposure with a slippery surfaces affects gait biomechanics. These studies have identified the *proactive strategies* that describe how people preemptively adjust their gait to reduce slip potential. Understanding how we reduce the potential for a slip to occur is important, as the large biomechanical changes required to recover after a slip can still lead to disabling injury even if a fall does not happen [3].

1.2.1.1. Kinematics and Kinetics

The ground reaction forces between the foot and floor have been implicated as being the most critical biomechanical factor in slip events [23]. The likelihood of a slip occurring can be quantified using the relationship between the required coefficient of friction and the available coefficient of friction. The required coefficient of friction is the ratio of the shear to normal forces generated during gait and must be lower than the friction available between the foot and floor for a slip to be avoided. As the difference between the required coefficient of friction and the available friction increase, so do the number of slips and falls, with this difference shown to predict the probability of a slip or fall event [24]. A logical step for researchers in this field was to determine how we reduce the required coefficient of friction during walking, to minimize the probability of a slip or fall event.

Several biomechanical factors have been implicated as important for deciding whether a slip results in recovery or a fall. Early work by Strandberg and Lanshammar showed if you allow your foot to slip beyond 10 cm or slide faster than 50 cm/s, it is likely you will fall [25], though subsequent work showed that these thresholds are not absolute [10, 11]. The implication of slip distance and sliding velocity as factors that determine the outcome of a slip highlights the importance of properly controlling shank and foot dynamics when interacting with slippery surfaces. One change that can facilitate more appropriate contact between the foot and ground is to reduce stride length, which has been implicated as a factor that determines the outcome of a fall [10, 13] and that contributes to reduced slip potential [26, 27]. A reduced stride length also reduces the braking and propulsive forces generated when walking [28], which reduces the required coefficient of friction and slip potential. In addition to reduced stride length, specific changes in ankle dynamics when stepping onto a slippery surface have been shown to influence the outcome of a slip or reduce slip potential. These include reducing the foot velocity when contacting the slippery surface, including the vertical velocity [12], horizontal velocity [4], and angular velocity [13, 26], and also reducing the foot-floor angle at heel contact [12, 13, 16, 26, 29]. From these studies it is clear that appropriately controlling your limb kinematics when interacting with a slippery surface is crucial for avoiding slip and fall events. Subsequent studies were interested in understanding the neural control of these changes in kinematics and resulting kinetics, which impedance describes the dynamic relationship between.

1.2.1.2. Muscle Activation

When considering the neural control employed when interacting with slippery surfaces, numerous studies found an increase in muscle activity across the lower-limb. In anticipation of stepping onto a possible slippery spot, people increased activity for muscles spanning the ankle [14, 16] and the knee [14] resulting in increased cocontraction. Chambers and Cham also found that when people naturally walked with increased ankle muscle cocontraction, they were more likely to experience a less severe slip and avoid a fall [14]. When walking continuously on a slippery walkway, Cappellini and colleagues showed an overall increase and broadening of lower-limb muscle activity throughout the gait cycle [15]. All of the cited studies concluded that this increase in muscle activation occurred to increase joint or limb stiffness to enhance stability on the slippery surface.

Surprisingly, there is evidence that suggests the observed increase in activity, specifically for the ankle muscles, is actually detrimental for avoiding slips and falls on slippery surfaces. While Marigold and Patla showed that when given knowledge of an impending slip, people increased medial gastrocnemius (ankle plantarflexor) activity for their first exposure to the slippery spot, in subsequent exposures as people continuously updated their reactive strategy, medial gastrocnemius activity was actually reduced [16]. After gaining experience with a slippery surface, Heiden and colleagues showed that tibialis anterior (ankle dorsiflexor) activity was reduced at heel-contact [29]. Lockhart and colleagues evaluated the strategy used when avoiding slipping on a single slippery spot after people had already experienced a slip on the same surface [27]. They showed when people attempted to avoid slipping, as lateral gastrocnemius (ankle plantarflexor) activity increased, so did the required coefficient of friction, which resulted in increased slip potential. They also showed as semitendinosus (knee flexor) activity increased, the required coefficient of friction decreased, indicating increased knee muscle activity might be an appropriate reaction for avoiding slips.

The observations of reduced ankle muscle activity highlight the importance of updating our strategy as we gain experience with a slippery surface to reduce slip potential for subsequent exposure. The results from Cappellini and colleagues, which found an overall increase in lower-limb muscle activity during initial exposure to a slippery walkway [15], further demonstrates the importance of updating this strategy. As subjects gained experienced with the slippery walkway, they were continuously slipping during mid-stance phase, where the increase in ankle muscle activity was observed. Subjects may not have updated their strategy or simply chosen a different strategy that involved controlled slipping on the slippery walkway. While the young adults who participated in the study did not fall while slipping on the slippery surface, it is unlikely that individuals with higher incidence of falls would have the same outcome, so it is necessary to identify the strategies that contribute to slip avoidance.

The majority of previous slipping studies analyzed how people interacted with a single slippery spot, and these interactions would likely be different if people instead had to take multiple steps on a slippery surface, which is common. Looking at the effect of a single slippery spot makes sense if you only consider encounters with an unexpected patch of ice. While these surfaces are incredibly dangerous, they are not particularly common. Much more commonly encountered, often on a daily basis, are moderately slippery surfaces like smooth hardwood floors or polished linoleum. These floors, especially without proper footwear, pose a real and constant danger, so understanding how people continuously interact with them without triggering a slip is crucial. A couple of studies have assessed how people walk on moderately slippery surfaces [30, 31], but neither evaluated the changes in lower-limb muscle activity, which have repeatedly been shown to be important for interacting with slippery surfaces. While numerous studies have asserted the importance of modulating lower-limb muscle activity for influencing limb stiffness, none of these studies evaluated the changes in muscle activity required to avoid slipping during continuous exposure to slippery surfaces. This presents a critical gap in this body of work in which we do not know how people adapt their gait to continuously avoid slipping on different types of slippery surfaces. In addition, the previous assertions that changes in muscle activity resulted in modulation of limb stiffness have not been experimentally tested. Given our current understanding of the modulation of impedance during movement, we cannot conclude with certainty that a change in muscle activation on a slippery walkway affects impedance.

1.2.2. Modulation of Joint Impedance

Studying joint impedance has several important applications including motor control research, understanding neuromuscular disease, and advancements in rehabilitation engineering [32]. For this reason, it has been studied extensively, but primarily under postural conditions when the factors known to influence impedance can be tightly controlled. Experimentally estimating joint impedance during movement is difficult for several reasons. It requires an apparatus and control system that allows the subject to move freely, while also having sufficient power to apply a perturbation to a single joint in isolation. It is also computationally challenging because traditional methods rely on assumptions of stationarity that do not hold during movement. Because of this, researchers have used some of the fundamental properties of impedance modulation identified during postural tasks to predict changes in impedance during movement. Whether these predictions are valid is unclear as our current understanding of the modulation of joint impedance during dynamic tasks is limited.

Under static conditions, impedance is dependent on the position of the joint and the level of muscle activation, which simultaneously change during movement. For a fixed ankle position, under relaxed conditions, Weiss and colleagues showed that stiffness, the static component of impedance, and damping, which represents the energy dissipating component of impedance, both increased for large dorsiflexion and plantarflexion angles [33]. During constant activation of the plantarflexors or dorsiflexors at fixed ankle positions, stiffness and damping increased from maximum plantarflexion to maximum dorsiflexion [7]. In addition, the relationship between muscle activity recorded through surface electromyography (EMG) and net joint torque was linear at all ankle angles, as was the relationship between stiffness and joint torque [7]. The dependence between muscle activation, joint torque, and stiffness persisted to maximum voluntary contraction, while damping remained essentially constant [8].

While the modulation of impedance during postural tasks is straightforward for joint position, joint torque, and muscle activation, their impact on impedance during movement is unclear. This is likely because studies evaluating impedance during movement were not designed to analyze the modulation of impedance while controlling for the other influential factors, as was done during the postural studies. Some studies have attempted to determine the effect of a change in either joint position or joint torque while the other was held constant. Even these more well-controlled studies have contradictory results.

Few studies have isolated the effect of a changing position on joint impedance and those that have exhibit mixed results. Kirsch and Kearney evaluated the effect of a large imposed stretch of the ankle plantarflexors (i.e. rapid dorsiflexion movement) on ankle stiffness [34]. They showed the stretch caused a large increase in torque that, despite some transient behavior in the estimated stiffness, was ultimately met with an increase in stiffness, as would be expected from the postural studies. This result opposes many studies that evaluated impedance during continuous movement, which consistently demonstrate a decrease in stiffness as a result of movement [19-21]. Specifically, Ludvig and Perreault showed when the knee muscles remained completely passive during an imposed sinusoidal movement, stiffness during the movement was significantly lower than that measured during relaxed isometric conditions [21]. This discrepancy between studies is likely due to differences in the imposed position change. Kirsch and Kearney imposed a rapid stretch that resulted in reflexive activity, while Ludvig and Perreault altered the position slowly and continuously, indicating the speed of the movement may also play a role in impedance modulation.

There have also been few studies that isolated the effect of changing joint torque on impedance. During a transient increase in ankle muscle activity, stiffness initially decreased before rapidly increasing to a higher steady value, while torque simply increased [35], demonstrating a more complex relationship than what is observed during posture. In contrast, a study evaluating the effect of dynamic muscle activation on knee stiffness showed when a single muscle group was activated (i.e. flexors or extensors), the increase is muscle activity was met with an increase in joint torque and stiffness [36]. The main difference in these studies was the time course in which stiffness was evaluated. In the former study, the transient changes in muscle activity and stiffness were captured over 100 ms, while the latter study averaged results over 100 ms windows, which could have smoothed out these transients.

The modulation of joint impedance during voluntary movement is more difficult to tease apart because the individual contributions of changing joint position and changing joint torque are still not clear. At the initiation of a single elbow flexion movement, a rapid burst of muscle activity to drive the movement resulted in increased torque and decreased stiffness [37]. Once the movement was completed, stiffness gradually increased to a steady value that was higher than before the movement occurred. In contrast, Popescu and colleagues showed for a single elbow extension movement, stiffness did not drop measurably during the movement, rather it was already low before the movement and remained low throughout [20]. When looking at continuous elbow movement, oscillating between flexion and extension, Bennett and colleagues showed the relationship between stiffness and net muscle torque was complex, increasing dramatically as the targets were reached and dropping steeply at the onset of the switch between flexion and extension [19]. The increase in complexity for the relationship between stiffness and joint torque was also observed during voluntary movement at the knee [21]. We can postulate that part of the unexplained behavior is caused by changing muscle activation, which we know produces some decoupling between stiffness and torque [35]. We can also postulate that the changing position contributed to the low stiffness values observed [21]. What remains unknown is the effect of simultaneously changing both joint position and joint torque because we do not know how changing one factor influences the relationship between the other factor and impedance. For example, we do not know how a changing joint position might further complicate the relationship between stiffness and torque during changing muscle activation, and vice versa. A well-controlled study that isolates the individual effect of a change in joint position and a change in joint torque, as well as, the effect of simultaneously changing them is needed.

1.2.3. Broader Significance

While this dissertation investigates the effect of altered neural control on joint impedance during dynamic conditions, including walking on a slippery surface, in unimpaired adults, this research has significant implications for populations for whom falls are more common. This includes older adults, for which it has been demonstrated a majority of fall-related hip fractures occurs on a wet or slippery surface [38], and individuals with a lower-limb amputation, who have previously cited a desire to be able to use their prosthesis on slippery surfaces [39].

Older adults were shown to naturally adopt a more cautious gait pattern, including a shorter stride length and smaller foot-floor angle at heel contact [13], which for a young adult would predispose them to experience a less hazardous slip. Despite this, older adults still experienced hazardous slips, slips likely to lead to a fall, at the same frequency as younger adults who exhibited less ideal gait patterns. It is unknown why these gait adaptations did not help older adults minimize slip severity and what strategies they should be adopting to facilitate less hazardous slips. In addition to studying why the reactive strategy used by older adults fails and results in a fall, [4, 13, 40], it may also be important to assess whether a failing proactive strategy is another contributor, resulting in more opportunities for their reactive strategy to fail. In this dissertation, we will evaluate the strategies that contribute to slip avoidance in young adults and with this information we can assess what is different in the strategies used by older adults and develop training programs to adjust their strategies accordingly.

The implication that joint mechanics are modulated by changes in muscle activity on the slippery surfaces is significant for both older adults and individuals with a lower-limb amputation. Muscle-tendon properties were shown to deteriorate with age, which contributed to a decline in postural stability during tasks more challenging than bipedal stance [41]. If joint mechanics are modulated on a slippery surface, the alteration in mechanics that occurs due to age may have a significant impact on an older adult's ability to adapt appropriately, contributing to a higher occurrence of slips and falls. Individuals with a lower-limb amputation are completely dependent on the capabilities of their prosthetic device, which currently do not allow for compensatory changes on different terrains. Recent advances in the development of biomimetic impedance-based control systems [42], are producing more naturally behaving prostheses [43] that automatically adapt to different terrains [44]. Integral to the development of these control systems is the knowledge of how the intact system modulates impedance on different terrains. By assessing successful slip avoidance strategies and determining whether these strategies involve a modulation of joint mechanics, will inform how advanced prosthetic devices should be controlled to provide safer ambulation for the user on slippery surfaces.

1.3. Specific Aims

Understanding the neural strategies employed and their mechanical consequences when walking on different terrains is essential for understanding why some interactions with variable terrains result in falls. Review of previous literature identified that the neural control employed when interacting with slippery surfaces has been implicated as having a mechanical consequence of changes in joint or limb stiffness. Deeper investigations into the literature revealed two critical gaps concerning our knowledge of walking on slippery surfaces and the link between neural strategies and their mechanical consequences.

Though a substantial amount of work has been done to understand how to minimize the likelihood of a fall following a slip, little work has been done to understand how the initial slip can be avoided altogether. When we think of interactions with slippery surfaces, it is likely we recall the catastrophic events that ensue after a slip and fall incident. In reality, it is much more likely that on a daily basis, we are continuously interacting with and adapting to low-friction surfaces in an attempt to avoid slipping without even realizing it. It is when these adaptations fail and a slip, and potentially a fall, is triggered when the event sticks in our minds. Understanding how we can decrease the opportunities for these slips to be triggered requires an understanding of what the appropriate adaptations are for slip avoidance.

Previous slipping studies highlighted the importance of changes in lower-limb muscle activity as a strategy for adapting the mechanics of the limb. Based on previous postural studies, it has been assumed that the increase in muscle activity seen during walking on slippery surfaces leads to an increase in ankle impedance. However, with our current understanding of the modulation of joint mechanics during movement, it is unclear if these assumptions are valid. Before we can definitively say whether the change in muscle activity seen during walking on slippery surfaces has a mechanical consequence, we must determine the effect of changes in neural activation on joint mechanics during dynamic conditions. The objective of this dissertation was to fill these gaps by investigating the following specifics aims.

Aim 1: Evaluate the changes in lower-limb muscle activity that occur when unimpaired people walk without slipping on different slippery surfaces.

We evaluated slip avoidance strategies in unimpaired adult humans as they walked, without slipping, on different types of slippery surfaces. This study addressed another gap in previous slipping research, which emphasized the use of extremely slippery surfaces, by analyzing the gait adaptations for both moderately and very slippery surfaces. The objective of this specific aim was to quantify the changes in lower-limb muscle activity that occurred when walking on the different slippery walkways and to discuss the potential consequences for joint mechanics. The working hypothesis was increasing slipperiness of the walkway would result in global increases in muscle activity across the lower-limb, likely indicating a stiffening of the lower-limb joints. This specific aim is discussed in Chapter 2.

Aim 2: Estimate unimpaired ankle impedance during steady state walking on a slippery walkway.

We estimated ankle impedance as unimpaired adult humans walked on a non-slippery walkway and a slippery walkway, without slipping. To estimate ankle impedance, we used a mechatronic platform that has been previously validated for use in estimating ankle impedance during the stance phase of gait [45]. This specific aim sought to test the hypothesis that the reduction in ankle muscle activity observed in Aim 1 would result in reduced shear forces and reduced ankle stiffness. This specific aim is discussed in Chapter 3.

Aim 3: Determine the individual and simultaneous effect of changing joint position and changing joint torque on stiffness.

While numerous studies have postulated the effects of a change in muscle activity on joint impedance during movement, no study has systematically addressed how movement impacts this relationship. During movement, we continuously alter the position of the joint as well as the joint torque that drives and brakes the movement. The objective of this specific aim was to determine how joint stiffness is affected by changes in joint position and changes in joint torque, when controlling for each factor. This specific aim is addressed in Chapter 4.

1.4. Dissertation Overview

Each chapter of the dissertation describes a single study that was completed to address one of the previously described specific aims. Chapter 2 is an article published in the IEEE Transactions on Biomedical Engineering investigating the gait characteristics that contribute to slip avoidance during continuous walking on a moderately and very slippery surface (Specific Aim 1). Chapter 3 is a manuscript that investigates the purpose of the observed reduction in ankle muscle activity on a slippery walkway, specifically testing the hypothesis that reduced muscle activity contributes to reduced shear forces and reduced ankle stiffness (Specific Aim 2). Chapter 4 is a manuscript that examines the modulation of joint impedance during movement, by determining how changing joint position and changing joint torque individually and in combination affect ankle stiffness (Specific Aim 3). Chapter 5 is a discussion of the dissertation work as a whole with concluding remarks on implications and future directions of this research.

2. Gait Characteristics When Walking on Different Slippery Walkways

2.1. Abstract

Objective: This study sought to determine the changes in muscle activity about the ankle, knee, and hip in able-bodied people walking at steady state on surfaces with different degrees of slipperiness. *Methods*: Muscle activity was measured through electromyographic signals from selected lower limb muscles and quantified to directly compare changes across surface conditions. *Results*: Our results showed distinct changes in the patterns of muscle activity controlling each joint. Muscles controlling the ankle showed a significant reduction in activity as the surface became more slippery, presumably resulting in a compliant distal joint to facilitate full contact with the surface. Select muscles about the knee and hip showed a significant increase in activity as the surface became more slippery. This resulted in increased knee and hip flexion likely contributing to a lowering of the body's center of mass and stabilization of the proximal leg and trunk. *Conclusion*: These findings suggest a proximal-distal gradient in the control of muscle activity that could inform the future design of adaptable prosthetic controllers. *Significance*: Walking on a slippery surface is extremely difficult, especially for individuals with lower limb amputations because current prostheses do not allow the compensatory changes in lower limb dynamics that occur involuntarily in unimpaired subjects. With recent advances in prosthetic control, there is the potential to provide some of these compensatory changes; however, we first need to understand how able-bodied individuals modulate their gait under these challenging conditions.

2.2. Introduction

The ability to change gait patterns when encountering different walking conditions is critical for minimizing the risk of falls. This is especially true when walking on a slippery surface, where falling carries a high risk of fracture [46]. Lower limb amputees are highly susceptible to falls; the prevalence of falls in this population may actually be higher than that for community-dwelling elderly people [47]. In addition, many lower limb amputees are also elderly [48], which compounds the likelihood of a fall. Lower limb amputees have indicated the importance of being able to ambulate on slippery surfaces [39], but the degree to which they can alter their gait on slippery surfaces is constrained by the inability of prosthetic limbs to adapt in the manner of an intact leg. Determining how able-bodied individuals adapt their legs to walk on slippery surfaces is the first step toward the design of lower limb prostheses that reduce the likelihood of slips.

Gait changes that occur when able-bodied people react to or anticipate stepping onto a single slippery spot have been studied extensively [14, 16, 23, 26, 29, 49]. With *a priori* knowledge, subjects proactively alter their gait dynamics to reduce slip potential. Increased lower limb muscle activity, including an increase in muscle cocontraction about the ankle and knee has been observed at heel contact [14, 15]. This increased cocontraction likely increases joint impedance and stability [50], an observation that is relevant to the development of biomimetic impedance-based control systems for lower limb prostheses [42] that promote more natural gait patterns [43]. When impedance parameters cannot be easily obtained, such as for walking on slippery surfaces, muscle activity, measured through electromyographic (EMG) signals, can be used as a first approximation of changes in joint mechanics [51].

Almost all previous slipping studies have focused on the changes occurring at the ankle and knee when interacting with discrete areas of extreme slipperiness. The steady-state changes that occur across the entire limb when walking continuously on a slippery walkway have not been completely quantified. One previous study investigated the initial adaptation to a simulated icy walkway [15]. They sought to understand how subjects adapt to a slippery walkway, rather than capturing the adapted changes that enable subjects to walk without slipping. As such, few trials were collected with gait speed minimally controlled and there was no quantitative comparison of EMG magnitude. Studying only extreme conditions resembling ice provides minimal information about how people walk on commonly encountered, moderately slippery surfaces, such as smooth hardwood or tile. These surfaces may not seem as dangerous, yet they are encountered more frequently than icy surfaces. While some studies have looked at gait on moderately slippery surfaces, including walking in socks on linoleum floors [31] and over a sheet of Teflon [30], to our knowledge no quantification of EMG or joint kinematics has been done on these surfaces. With most slipping studies using a simulated icy surface, it is interesting that minimal work has been done on changes at the hip, because often the most perceptible result of walking over ice is subsequent soreness in the hips. The main study that considered hip contributions to walking on an extremely slippery surface reported that lateral hip oscillations were reduced as subjects adapted to the conditions, but did not address the changes in muscle activity [15]. Another study that monitored hip muscle activity did so during reactions to unexpected slips, rather than capturing the proactive changes that aid in avoiding a slip [40]. The relative contribution of the hip muscles to safe ambulation on slippery surfaces is important to understand, as the majority of lower limb amputations occur at the transtibial (above ankle) or transfemoral (above knee) level [52], thus hip muscles are available in most lower limb amputees.

The purpose of this study was to quantify the activation of lower limb muscles during steady-state able-bodied walking over walkways with differing degrees of slipperiness. Based on previous work [15], we hypothesized that increasing slipperiness of the walkway would result in global increases in muscle activity about the ankle, knee, and hip, likely indicating an overall stiffening of the lower limb joints. This hypothesis was tested by recording EMG from lower limb muscles as subjects walked across a nonslippery walkway and two walkways of different slipperiness. Changes in muscle activity were quantified to estimate how joint mechanics are modified to enable walking on slippery surfaces. We believe that our results have important implications for the future design of prosthetic legs that can adapt to locomotion on different surfaces. A preliminary version of this study has been reported [53].

2.3. Methods

2.3.1. Walkway Surfaces

In previous slipping studies, lubricants were used to mimic very slippery surfaces like ice [54]. A goal of this study was to understand how people walk on surfaces that are moderately slippery but, because they are so common, still pose a substantial hazard. To achieve this goal, a novel walkway was created using high-gloss, laminated flooring treated with furniture polish to create a slippery surface that could easily be found in the home.

We characterized how people walk on three different 6-m walkways with different levels of slipperiness (see Figure 2.1). The coefficient of friction (COF) of each walkway was determined using an inclined sled test. The test identified the angle at which the foot interface began to slide over the walkway surface, which is directly related to the static COF. The following walkways

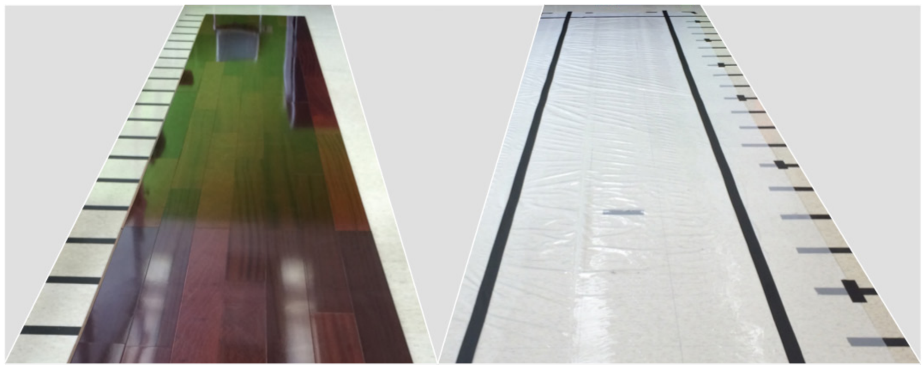


Figure 2.1. Walkway Surfaces. Laminate flooring for NS and MS walkways (left) and lubricated plastic sheeting for VS walkway (right).

were used.

1) *Nonslippery (NS)*: This was used as a baseline for comparison to the slippery walkways. Subjects walked across polished laminate flooring wearing treaded socks, which provided grip for a higher friction surface (COF > 0.4).

2) *Moderately Slippery (MS)*: This was used to capture how people walk on a surface that is less slippery than ice. Subjects wore socks as they walked across the polished laminate flooring (COF = 0.17 ± 0.01).

3) *Very Slippery (VS)*: This walkway was used to capture how subjects walk on a surface that is as slippery as ice. As for other slipping studies [15], subjects wore plastic booties over their bare feet and walked across a sheet of plastic covered in mineral oil (COF = 0.08 ± 0.01).

2.3.2. Protocol

Eleven able-bodied subjects (four males, seven females; 28 ± 4 years, 66 ± 16 kg, 172 ± 11 cm) gave written informed consent to participate in this study, which was approved by the Northwestern University Institutional Review Board. Each subject wore a safety harness fastened to an overhead gantry system to catch the subject in the event of a fall, though no falls occurred in this study.

For each walkway, subjects were instructed to walk across the entire 6 m at a self-selected comfortable pace. They did this 20 times; each repetition is considered a *trial*. As would be expected [15], subjects walked at different paces across each of the test surfaces. Therefore, after completing these self-selected pace trials, we collected a set of controlled-pace trials to separate changes in walking strategy that resulted from a change in pace from those due only to the change in surface type. In these trials, subjects were instructed to walk at a pace of 75 steps/min, assisted by a metronome. This pace was selected because it was just below the lowest pace observed in a set of preliminary trials assessing self-selected walking paces across all surfaces. Again, 20 trials were collected for each surface.

All subjects began on the NS walkway to capture baseline gait patterns in NS conditions. The order in which subjects experienced the slippery walkways was randomized to account for potential order effects associated with adaptation to the slippery surfaces [15]. Finally, all subjects repeated the NS walkway at the end of experiment to further control for learning during the course of the experiment. For subsequent analysis and discussion, the first 20 trials recorded on the NS walkway will be referred to as NS1 and the second 20 trials at the end of the experiment will be referred to as NS2. Specifically, five subjects completed the walkway trials in the order: NS1 → MS → VS → NS2. The remaining six subjects completed the trials in the order: NS1 → VS → MS → NS2. The entire experiment took about 1 h and rest breaks were taken after completion of each walkway.

EMG signals were collected using bipolar surface electrodes (model DE2.1; Delsys, Boston, MA, USA) from 11 muscles in the right leg: medial gastrocnemius (MG), lateral gastrocnemius (LG), soleus (SOL), tibialis anterior (TA), vastus medialis (VM), vastus lateralis (VL), semitendinosus (ST), biceps femoris (BF), adductor longus (AL), gluteus maximus (Gmax), and gluteus medius (Gmed). These muscles represent both uniarticular and biarticular muscles about the ankle, knee, and hip. Placement of surface electrodes was guided using SENIAM standards and verified by visualizing EMG activity generated during test contractions. The signals were amplified 10 000×, filtered between 20 and 450 Hz (Bagnoli 16, Delsys), and sampled at 1 kHz. Kinematic data were collected using biaxial goniometers attached at the ankle, knee, and hip (Biometrics, Ltd.) and sampled at 500 Hz; kinematic data were only available for ten of the 11 subjects. Subjects were also outfitted with two custom-made footswitches taped to the sole of their right foot at the heel and the big toe. Footswitch data were sampled at 1 kHz and were used to line up strides from individual gait trials. All signals were acquired simultaneously using a custom 16-bit analog-to-digital converter that allows for multirate sampling.

2.3.3. Data Analysis

The first five trials from both the self-selected and speed-controlled datasets were omitted from the analysis to account for subjects becoming accustomed to the surface conditions and to walking to the metronome. Each of the remaining 15 gait trials was separated into individual strides. To avoid gait initiation and termination effects, the first and last full strides were omitted from the analysis [55, 56]. This resulted in approximately 30–60 strides per subject for each walkway. The exact number of strides depended on the walkway. For example, there tended to be approximately 30 strides for the NS walkway, but closer to 60 strides for the VS walkway due to the shorter step length often chosen for these more slippery conditions (see Table 2.1). The kinematic waveforms for all strides were visually confirmed for consistency to ensure that no

Table 2.1. Average pace, speed, and step length. (Mean ± S.E) N = 10

Walkway	Self-Selected Pace (steps/min)	Control Trials: Pace (steps/min)	Self-Selected Speed (m/min)	Control Trials: Speed (m/min)	Self-Selected Step Length (m/step)	Control Trials: Step Length (m/step)
NS1	110.6 ± 0.7	77.8 ± 0.2	69.8 ± 0.5	42.9 ± 0.2	0.63 ± 0.01	0.55 ± 0.01
NS2	106.8 ± 0.7	77.2 ± 0.2	65.0 ± 0.5	44.1 ± 0.2	0.61 ± 0.01	0.57 ± 0.01
MS	105.3 ± 0.7	77.4 ± 0.2	59.3 ± 0.5	41.5 ± 0.2	0.56 ± 0.01	0.54 ± 0.01
VS	93.2 ± 0.7	76.4 ± 0.2	46.0 ± 0.5	37.3 ± 0.2	0.50 ± 0.01	0.49 ± 0.01

additional initiation or termination strides were included. The remaining strides were divided into stance phase and swing phase.

EMG signals were notch filtered at 60 Hz, full-wave rectified, and normalized in time to allow comparisons across walking speeds. This was accomplished using linear interpolation to obtain 600 points uniformly spaced in time during stance phase and 400 points uniformly spaced in time during swing phase. These values were chosen based on the typical stride, which is 60% stance and 40% swing [57]. To normalize EMG magnitude, a moving average with a 0.5-s window was applied to every stride for all walkways, including both self-selected and controlled-speed strides. The maximum of every stride was taken and the overall maximum value was used for normalization. A zero-lag 50 ms root-mean-square (RMS) filter was used to visualize EMG and to quantify cocontractions, as described below.

EMG amplitude was quantified by the RMS value over different windows throughout stance and swing phases. These included 0%–100% stance (overall stance), 0%–17% stance (initial contact), 17%–83% stance (mid-stance), 83%–100% stance (pre-swing), 0%–100% swing (overall swing), 0%–50% swing (early swing), and 50%–100% swing (late swing) [57]. Cocontraction between antagonistic muscle pairs was quantified at each joint using a representative muscle when multiple muscles that performed the same action were recorded (i.e., MG was chosen as the representative ankle plantarflexor). These included MG and TA (ankle), VL and ST (knee), AL and Gmax (hip flexion/extension), and Gmed and Gmax (hip abduction/adduction). For each pair, cocontraction was quantified by the cocontraction index (CCI) as defined by Rudolph *et al.* [58]:

$$\mathrm{CCI} = \frac{\mathrm{LowEMG_i}}{\mathrm{HighEMG_i}} * (\mathrm{LowEMG_i} + \mathrm{HighEMG_i}) \quad (1)$$

Low EMG refers to the activity of the less active muscle of the pair, whereas High EMG refers to the activity of the more active muscle. The CCI was computed at each time point within stance phase and swing phase.

Video data were used to detect slipping, which only occurred on the VS walkway. For the self-selected trials, slips were only common within the first five trials that were already omitted from the analysis. For the remaining 15 trials considered in the analysis, slips were rare for steady-state strides during both the self-selected and controlled-speed trials (i.e., on average 1–2 slips occurred per subject over all subsequent trials). Strides containing slips were omitted from the analysis. Video data were also used to estimate walking speed, which were fully available for ten of the 11 tested subjects (one subject had missing video data for NS2).

2.3.4. Statistics

We hypothesized that all lower limb muscle activity would increase as the walkway became more slippery. A linear mixed-effects model was used to test for significant differences in EMG amplitude across walkways. The fixed factor of the model was walkway (NS1, MS, VS, NS2), with subject treated as a random factor. The hypothesis was tested in each of the stance and swing phase windows described before. Results for the window encompassing the entire stance or swing period are presented, except when the results from the smaller windows were substantially different. This same statistical model was also used to assess significant differences in joint angles and the CCI at every 5% of the stride, and average pace, speed, and step length across all recorded strides. Post hoc comparisons were done using Bonferroni corrections, testing for significance at a 5% level.

2.4. Results

Significant differences in lower limb muscle activity and gait characteristics were observed about the ankle, knee, and hip when walking on slippery and NS walkways for both the self-selected and controlled-speed trials. We first present the results for the self-selected trials in full, and then present the results for the controlled-speed trials that differed from the self-selected trials. There were minimal differences in kinematics between the self-selected and controlled-speed trials, none of which affected the trends observed and presented, so the differences in EMG are emphasized. Although many of the lower limb muscles chosen for analysis are biarticular, for simplicity, we describe each muscle according to the individual joint at which its actions are the greatest, as defined by the SENIAM guidelines [59]. In addition, within the group of recorded muscles existed sets of muscles that produced the same major action. These included MG, LG, and SOL as ankle plantarflexors, VL and VM as knee extensors, and ST and BF as knee flexors. The resulting muscle activity in each set was highly similar (on average 90% of the post hoc testing was the same across muscles). As such, a representative muscle from each set was chosen to be presented.

2.4.1. Pace and Speed

Subjects gradually reduced their pace, speed, and step length as the walkway became more slippery during the self-selected trials, whereas a more consistent pace and speed were achieved during the controlled-speed trials. There was a significant effect of walkway for the self-selected pace ($F_{3,1552} = 301.2$; $p < 0.001$), speed ($F_{3,1552} = 1052$; $p < 0.001$), and step length ($F_{3,633} = 217$; $p < 0.001$), in which NS1 > MS > VS ($p < 0.001$ all comparisons; Table 2.1). Self-selected pace, speed, and step length were also reduced on NS2 when compared to NS1 ($p < 0.001$ all comparisons).

Although subjects closely matched the prescribed metronome pace during the controlled-speed trials on all walkways, minor yet statistically significant differences emerged between speeds across walkways. On average, subjects walked at a slightly higher pace than the 75 beats/min specified by the metronome (see Table 2.1, Column 2). There was a significant effect of walkway for pace ($F_{3,1663} = 19.8$; $p < 0.001$), with subjects adopting a significantly slower pace on VS compared to the other walkways ($p < 0.001$ all comparisons). There was also a significant effect of walkway for speed ($F_{3,1663} = 477$; $p < 0.001$) and step length ($F_{3,628} = 68.4$; $p < 0.001$). Significant differences were observed between nearly all walkways for speed and step length ($p < 0.01$ all comparisons), with the exception being the comparison between step length on NS1 and MS ($p = 0.34$). Subjects adopted a significantly shorter step length on VS compared to NS1 and NS2 ($p < 0.01$), which resulted in a slower speed ($p < 0.001$). Finally, subjects adopted a larger step length on NS2 compared to NS1 ($p = 0.007$), which resulted in a significantly higher speed on NS2 ($p < 0.001$). Overall, the range of speeds achieved during the controlled-speed trials was reduced by over 70% when compared to the self-selected trials, providing a dataset in which the effect of speed on changes in EMG and kinematics was minimized.

2.4.2. Ankle

MG (an ankle plantarflexor) and TA (an ankle dorsiflexor) showed reduced activity on the slippery walkways during the period of stance in which they are typically active and opposing changes during swing. For a representative subject, there were large differences in MG activity during midstance [see Figure 2.2(a) left-shaded area], in which activity was highest on NS1 and lowest on VS. This was consistent across subjects [see Figure 2.2(c) left side], where a significant effect of walkway was present ($F_{3,1737} = 269$; $p < 0.001$). Post hoc testing revealed that muscle activity was gradually reduced as the walkway became more slippery with NS1 > MS > VS ($p < 0.001$ all comparisons). Activity was also reduced during the NS2 trials compared to NS1 ($p < 0.001$), but was still larger than MS and VS ($p < 0.001$). During late swing, an opposite trend emerged when looking across subjects [see Figure 2.2(c) right side], but was weakly present, if at all, in the representative subject [see Figure 2.2(a) right-shaded area]. There was a significant effect of walkway ($F_{3,1737} = 88.6$; $p < 0.001$) with post hoc testing revealing that muscle activity was

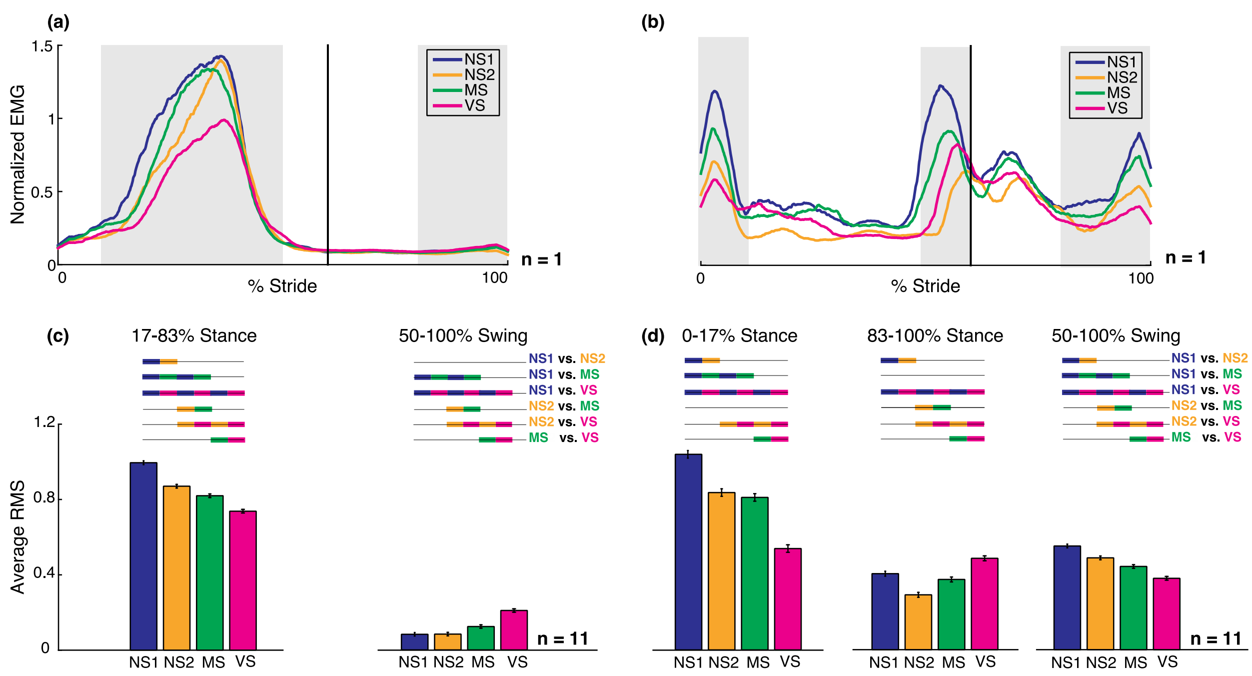


Figure 2.2. Influence of walkway surface on ankle EMG. Representative EMG for a single subject for (a) MG and (b) TA. Gray areas indicate windows of EMG quantified using RMS with results shown for (c) MG during midstance and late swing (n = 11) and (d) TA during initial contact, preswing, and late swing (n = 11). Shading indicates $p < 0.05$. Error bars indicate S.E.

gradually increased as the walkway became more slippery, VS > MS > NS1 and NS2 ($p < 0.002$ all comparisons). TA showed the strongest activity during initial contact with the representative subject showing highest activity on NS1 and lowest on VS [see Figure 2.2(b) left-shaded area]. This was consistent across subjects with a significant effect of walkway [see Figure 2.2(d) left side; $F_{3,1737} = 226$; $p < 0.001$]. Post hoc testing revealed that muscle activity was gradually reduced on the slippery walkways with NS1 > MS > VS ($p < 0.001$ all comparisons). Additionally, muscle activity was lower on NS2 compared to NS1 ($p < 0.001$). During preswing, differences were less consistent across subjects, but a significant effect of walkway was found [see Figure 2.2(d) middle; $F_{3,1737} = 77.1$; $p < 0.001$]. Post hoc testing revealed muscle activity was highest on VS ($p < 0.001$ all comparisons) and lowest on NS2 ($p < 0.001$ all comparisons), yet only the latter appears obvious in the representative subject [see Figure 2.2(b) middle-shaded area]. Finally, during late swing [see Figure 2.2(b) and (d) right side], there was a significant effect of walkway ($F_{3,1737} = 107$; $p < 0.001$) with muscle activity once again gradually reduced as the walkway became more slippery, NS1 > NS2 > MS > VS ($p < 0.001$ all comparisons).

A couple of key differences were observed between walkways for the CCI at the ankle. At heel contact [0% stride, Figure 2.3(a)], there was a small increase in CCI on VS ($p < 0.001$ all comparisons). During the majority of midstance (10%–50% stride), CCI was highest on NS1 when compared to NS2 and MS ($p < 0.03$ for shaded areas indicating significance), but only greater than VS toward the end of midstance (40%–50% stride, $p < 0.001$ all comparisons). Finally, during late swing (80%–100% stride), CCI was highest on VS when compared to all other walkways ($p < 0.001$ all comparisons).

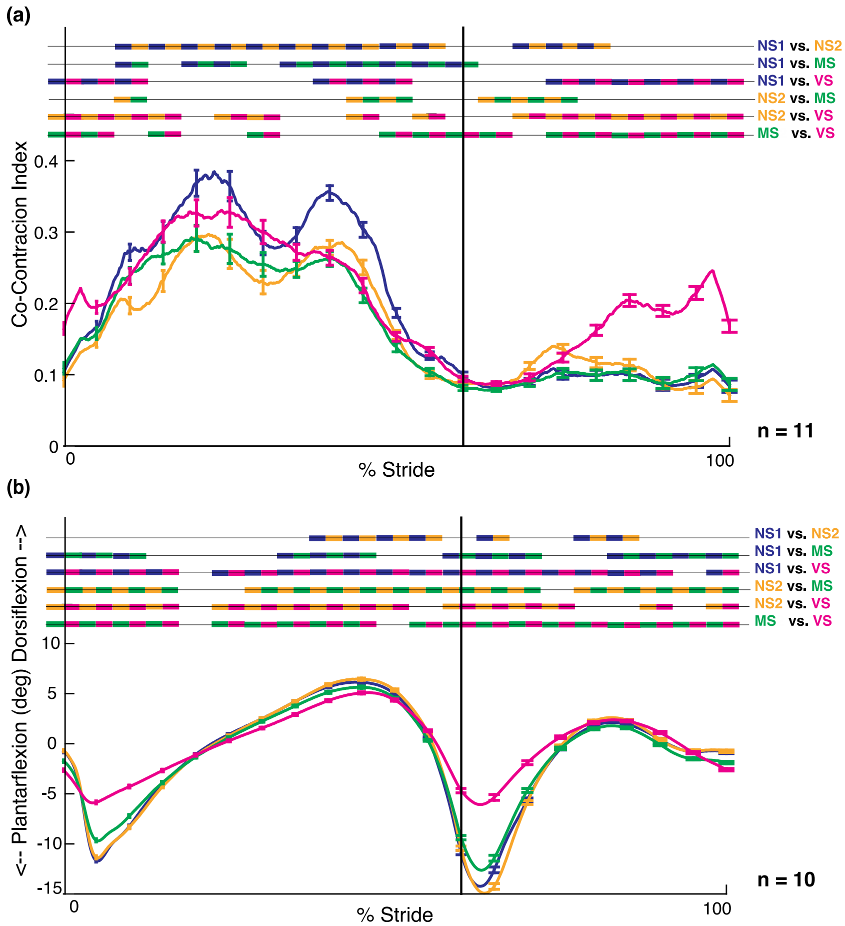


Figure 2.3. Influence of walkway surface on ankle CCI and kinematics. (a) Average degree of cocontraction for MG versus TA (n = 11) and (b) average ankle angle (n = 10). Shading indicates $p < 0.05$. Error bars indicate S.E.

Gradual changes in the ankle angle occurred as the walkway became more slippery, resulting in a reduced range of ankle angle during the stride. At heel contact [0%, Figure 2.3(b)], ankle angle was gradually reduced on the slippery walkways with NS1 and NS2 > MS > VS ($p < 0.001$ all comparisons). Additionally, when the ankle angle typically reaches maximum plantarflexion during stance phase (5% stride), the degree of plantarflexion was gradually reduced as the walkway became more slippery, NS1 and NS2 > MS > VS ($p < 0.001$ all comparisons). Thereafter, as the ankle angle moves toward maximum dorsiflexion during stance, there was reduced dorsiflexion on the VS walkway starting at 25% stride ($p < 0.001$ all comparisons) and reduced dorsiflexion on MS compared to NS1 and NS2 starting at 35% stride ($p < 0.001$). During the transition to swing phase (60%–70% stride), when the ankle angle is once again biased toward plantarflexion, there was a gradual decrease in plantarflexion as the walkway became more slippery, NS1 and NS2 > MS > VS ($p < 0.001$ all comparisons). Finally, swing phase also ends with the ankle angle gradually reduced, NS1 and NS2 > MS > VS ($p < 0.001$ all comparisons).

2.4.3. Knee

Knee muscle activity showed less consistent trends as the walkway became more slippery than the ankle muscles. VL, a knee extensor, showed only a few consistent changes across subjects, where for a representative subject [see Figure 2.4(a) left-shaded area], there appeared to be some interesting effects of the walkway surface throughout stance, but these effects were not present across subjects. For overall stance, there was a significant effect of walkway on the VL stance phase EMG [see Figure 2.4(c) left side; $F_{3,1737} = 37.7$; $p < 0.001$], with muscle activity lowest on NS2 ($p < 0.001$ all comparisons). During late swing, a more consistent trend emerged across subjects, demonstrated by the representative subject [see Figure 2.4(a) right-shaded area], in which activity was highest on NS1 and lowest on VS. During this period, there was a significant effect of walkway [see Figure 2.4(c) right side; $F_{3,1737} = 20.9$; $p < 0.001$] with post hoc testing revealing that muscle activity was gradually reduced as the walkway became more slippery, NS1 > MS > VS ($p < 0.004$ all comparisons). Muscle activity was also reduced on NS2 compared to NS1 ($p < 0.001$). ST, a knee flexor, showed strong activity during initial contact [see Figure 2.4(b) and (d), left side], where there was a significant effect of walkway ($F_{3,1737} = 122$; $p < 0.001$). Muscle activity

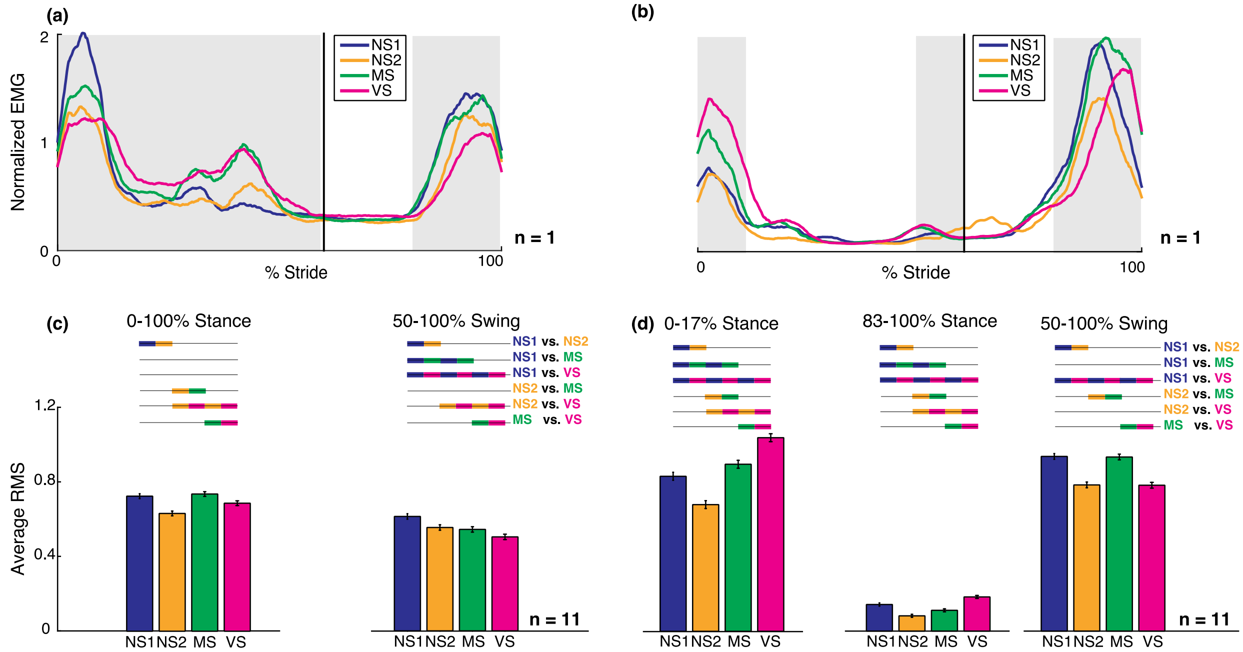


Figure 2.4. Influence of walkway surface on knee EMG. Representative EMG for a single subject for (a) VL and (b) ST. Gray areas indicate the windows of EMG quantified using RMS with results shown for (c) VL for overall stance and late swing (n = 11) and (d) ST during initial contact, preswing, and late swing (n = 11). Shading indicates $p < 0.05$. Error bars indicate S.E.

gradually increased as the walkway became more slippery, VS > MS > NS1 and NS2 ($p < 0.01$ all comparisons), which can be clearly seen in the representative subject EMG [see Figure 2.4(b) left-shaded area]. Additionally, muscle activity was reduced on NS2 compared to NS1 ($p < 0.001$). During preswing [see Figure 2.4(d) middle], there was also a significant effect of walkway ($F_{3,1737} = 72.3$; $p < 0.001$). Activity was highest on VS ($p < 0.001$ all comparisons), yet the effect was small as shown by the representative subject [see Figure 2.4(b) middle-shaded area]. Finally, during late swing [see Figure 2.4(b) and (d) right side], there was a significant effect of walkway ($F_{3,1737} = 69.6$; $p < 0.001$), yet no consistent trend emerged between the walkways with activity lowest on VS and NS2 ($p < 0.001$ all comparisons).

CCI at the knee was increased on VS during portions of stance phase, but reduced on the slippery walkways for late swing phase. From 10% to 25% stride and 50% to 60% stride [see Figure 2.5(a)], there was an increase in CCI on VS ($p < 0.001$ all comparisons). During late swing phase (85%–90% stride), there was a gradual decrease in CCI as the walkway became more slippery with NS1 > MS > VS ($p < 0.02$ all comparisons), and VS persisting to have the lowest CCI until the end of the stride ($p < 0.001$ all comparisons).

Gradual changes in the knee angle occurred as the walkway became more slippery that resulted in a smaller range in knee angle achieved throughout the stride. At heel contact [see Figure 2.5(b), 0% stride], there was a gradual increase in knee flexion with VS > MS > NS1 and NS2 ($p< 0.03$ all comparisons). Increased knee flexion lasted on VS through most of stance phase until

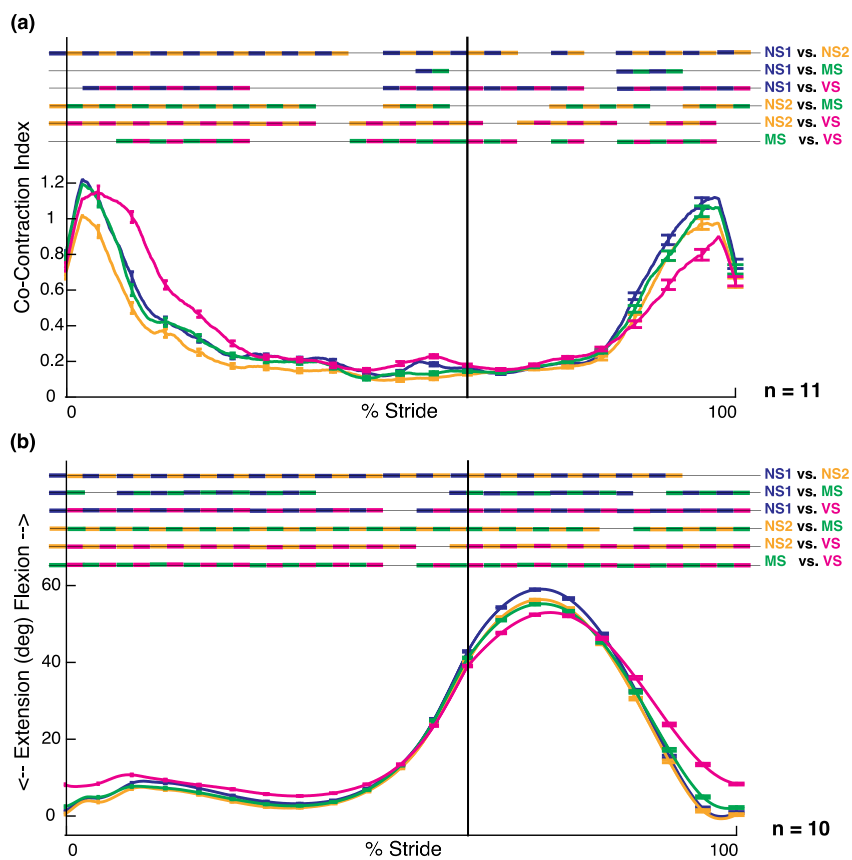


Figure 2.5. Influence of walkway surface on knee CCI and kinematics. (a) Average degree of cocontraction for VL versus ST (n = 11) and (b) average knee angle (n = 10). Shading indicates $p < 0.05$. Error bars indicate S.E.

50% stride ($p < 0.001$ all comparisons). There was also increased knee flexion on MS compared to NS2 throughout all of stance ($p < 0.01$ all comparisons); though statistically significant, these differences were small. From the transition to swing phase and throughout early swing (60%–75% stride), there was a gradual decrease in knee flexion as the walkway became more slippery with NS1 and NS2 > MS > VS ($p < 0.003$ all comparisons). Thereafter activity reversed, and the degree of knee extension achieved was gradually reduced as the walkway became more slippery, NS1 and NS2 > MS > VS ($p < 0.001$ all comparisons).

2.4.4. Hip

The measured hip muscles showed variable changes in muscle activity, with some muscles showing increased or decreased activity on the slippery walkways. AL, a hip adductor and flexor, showed only a few consistent changes across subjects, but there was still a significant effect of walkway [see Figure 2.6(d) left side; $F_{3,1241} = 19.0$; $p < 0.001$]. Post hoc comparisons revealed that muscle activity was increased on MS and VS when compared to NS2 ($p < 0.001$), but only MS was greater than NS1 ($p < 0.001$). These effects were small and are not obvious in the representative subject EMG [see Figure 2.6(a) left-shaded area]. Overall swing (0%–100% swing) is also presented for AL [see Figure 2.6(d) right side] where there was a significant effect of walkway ($F_{3,1241} = 19.3$; $p < 0.001$). Here, post hoc testing revealed that both MS and VS were greater than NS1 and NS2 ($p < 0.001$ all comparisons), but only an increase in MS is obvious in the representative subject [see Figure 2.6(a) right-shaded area]. Overall stance and swing are also presented for Gmax, a hip adductor and extensor [see Figure 2.6(b) and (e)]. There was a significant effect of walkway for overall stance ($F_{3,1454} = 32.1$; $p < 0.001$) and for overall swing ($F_{3,1454} = 26.1$; $p < 0.001$). Muscle activity was highest on NS1 compared to all other walkways (p

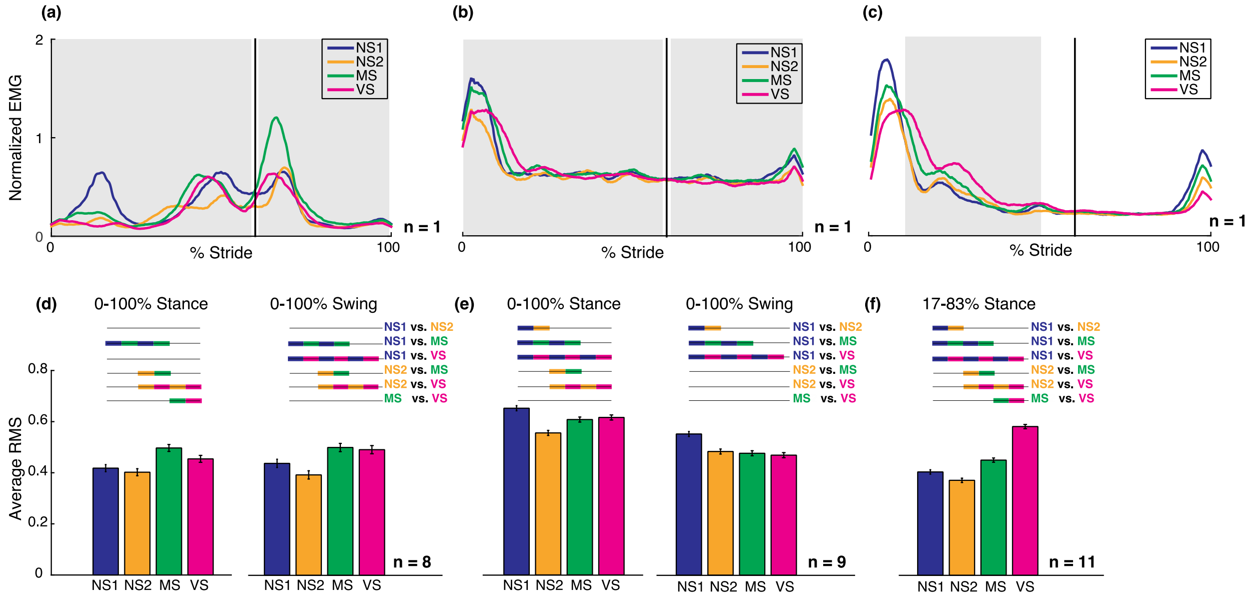


Figure 2.6. Influence of walkway surface on hip EMG. Representative EMG for a single subject for (a) AL, (b) Gmax, and (c) Gmed. Gray areas indicate windows of EMG quantified using RMS with results for (d) AL for overall stance and overall swing (n = 8), (e) Gmax for overall stance and overall swing (n = 9), and (f) Gmed for midstance (n = 11). Shading indicates p < 0.05. Error bars indicate S.E.

< 0.005 all comparisons). Finally, Gmed, a hip abductor, showed a significant effect of walkway for midstance [see Figure 2.6(c) and (f); $F_{3,1737} = 305$; $p < 0.001$]. Activity was gradually increased on the slippery walkways with VS > MS > NS1 and NS2 ($p < 0.001$ all comparisons). In addition, activity was reduced on NS2 when compared to NS1 ($p = 0.02$).

The CCI at the hip was enhanced during initial contact on the VS walkway. All three hip muscles assist in more than one degree of freedom about the hip, so isolating CCI in one degree of freedom is difficult. Nevertheless, we observed a consistent trend toward increased CCI on VS compared to all other walkways for 5%–10% stride between AL and Gmax [see Figure 2.7(a), p <0.002] and for 10%–45% stride between Gmax and Gmed [see Figure 2.7(b), $p < 0.001$]. Swing phase in general showed minimal significant differences between walkways.

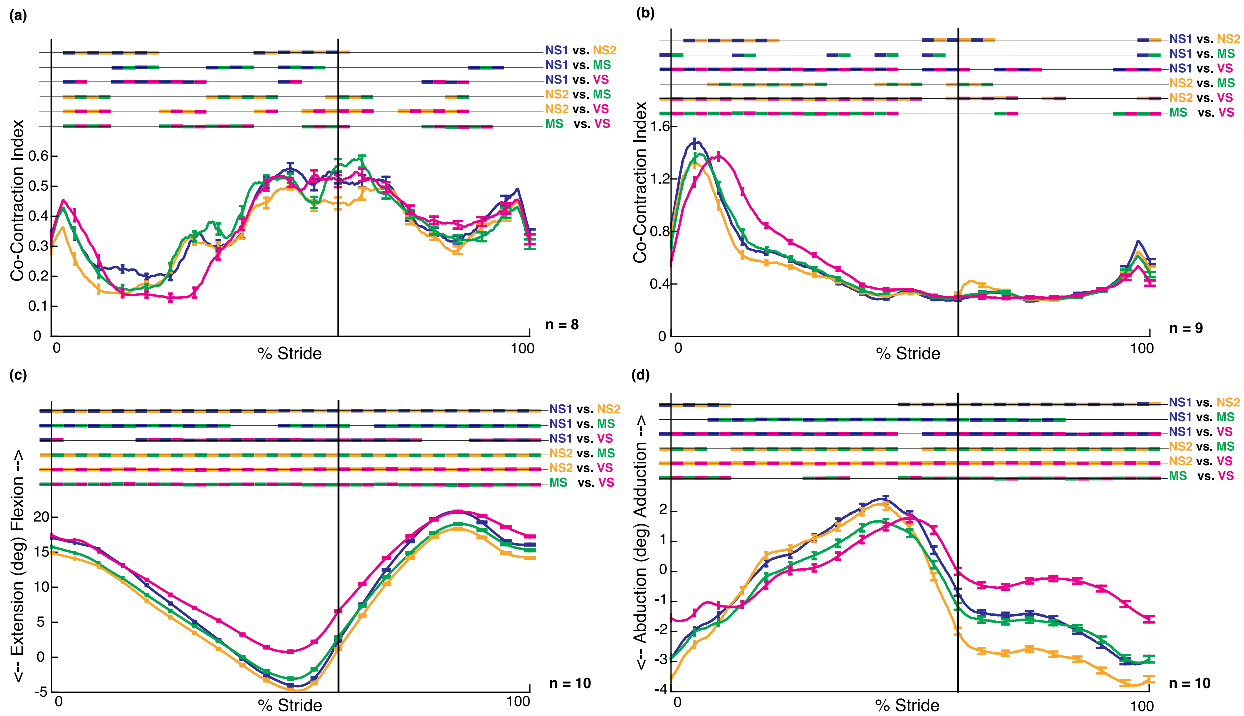


Figure 2.7. Influence of walkway surface on hip CCI and kinematics. (a) Average CCI for AL versus Gmax about hip flexion/extension (n = 8) and (b) average CCI for Gmed versus Gmax about hip abduction/adductino (n = 9). (c) Average hip extension/flexion angle (n = 10) and (h) average hip abduction/adduction angle (n = 10). Shaded areas indicate $p < 0.05$. Error bars indicate S.E.

Distinct trends were observed for hip extension/flexion (sagittal plane motion) and hip abduction/adduction (frontal plane motion) on the slippery walkways, with constant changes throughout all of stance in the sagittal plane and more time varying behavior in the frontal plane. In the sagittal plane [see Figure 2.7(c)], hip flexion was greatest on VS starting at 20% stride lasting throughout nearly the entire stride ($p < 0.003$ for shaded areas indicating significance). Hip flexion was also lowest on NS2 for the entire stride ($p < 0.003$ all comparisons). In the frontal plane [see Figure 2.7(d)], the stride started with adduction highest on VS ($p < 0.001$ all comparisons). Starting after 15% stride, the degree of adduction was reduced on the slippery walkways such that NS1 and NS2 > MS and VS until 50% stride ($p < 0.03$ all comparisons).

During the transition to swing phase, there was once again enhanced adduction on VS that lasted through the end of the stride ($p < 0.001$ all comparisons). Additionally, during swing phase, there was greatly reduced adduction on NS2 compared to all other walkways ($p < 0.001$ all comparisons).

2.4.5. Controlled-Speed Trials

Most comparisons between muscles were similar between the self-selected and controlled-speed trials, including MG, TA, ST, AL, and Gmed. On average, more than 75% of the post hoc testing done for these muscles yielded similar conclusions between the self-selected and controlled-speed conditions. An example of the striking similarity between the self-selected and controlled-speed trials can be seen for MG in Figure 2.8(a) and (d), compared to the self-selected trials shown in Figure 2.2(a) and (c). The post hoc testing for midstance was identical between the two datasets, but during late swing, there was no longer an increase in activity on MS relative to NS1 and NS2 ($p > 0.05$).

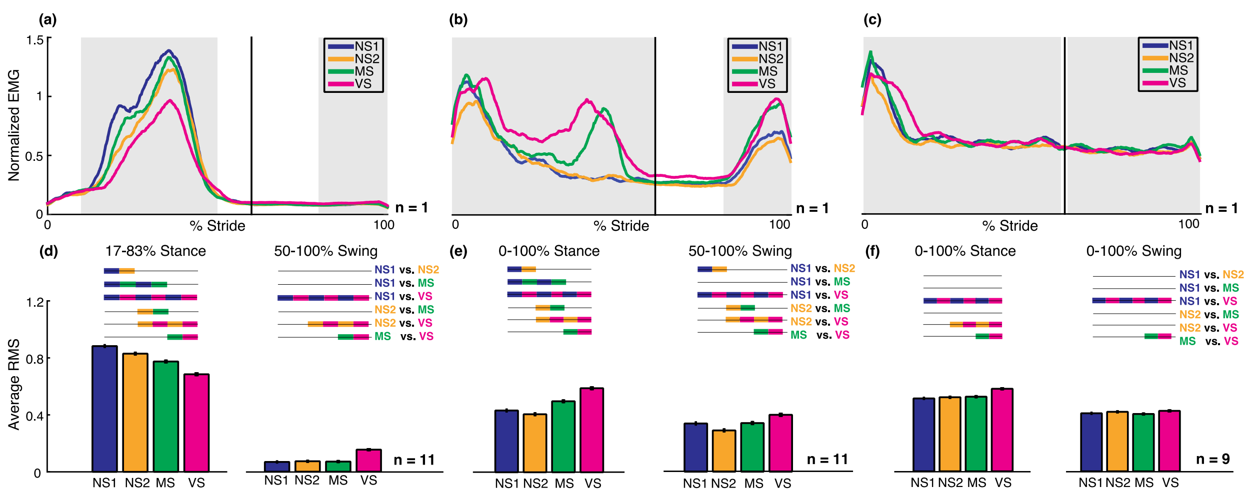


Figure 2.8. EMG quantification of select muscles for the controlled-speed trials. Representative EMG for a single subject is shown for (a) MG, (b) VL, and (c) Gmax. Gray areas indicate the windows of EMG quantified using RMS with results for (d) MG for midstance and late swing (n = 11), (e) VL for overall stance and late swing (n = 11) and (f) Gmax for overall stance and overall swing (n = 9). Shading indicates $p < 0.05$. Error bars indicate S.E.

There were some notable differences between the EMG quantification for the self-selected and controlled-speed trials, specifically for VL and Gmax. Both muscles still showed a significant main effect of walkway for the controlled-speed trials, but post hoc comparisons differed. For the controlled-speed trials, VL had gradually increasing muscle activity as the walkway became more slippery [see Figure 2.8(b) and (e)]. This was present during overall stance, with VS > MS > NS1 and NS2 ($p < 0.001$ all comparisons) and nearly present during late swing, with VS > MS, NS1, and NS2 ($p < 0.001$ all comparisons), but MS only greater than NS2 ($p < 0.001$). Muscle activity for Gmax also presented differently in the controlled-speed trials, where activity was generally highest on VS [$p < 0.02$; Figure 2.8(c) and (f)], as compared to the self-selected trials in which activity was highest on NS1 [see Figure 2.6(e)].

2.5. Discussion

The purpose of this study was to quantify the changes in lower limb muscle activity that occurred as people walked at steady state on various slippery surfaces. We hypothesized that there would be a global increase in muscle activity as the walkway became more slippery, thereby contributing to increased limb impedance and stability. The results of this study indicate that walking on slippery surfaces requires more complex control than originally hypothesized. Specifically, distinct changes occurred at each joint, and the results for the ankle during stance phase were in direct disagreement with the original hypothesis.

2.5.1. Proximal to Distal Control

The recorded ankle muscles had reduced activity on the slippery walkways during the period of stance in which they are typically active. This reduction was accompanied by systematic changes in the ankle kinematics, and likely contributed to decreased impedance of the joint. At heel contact, ankle dorsiflexor (TA) activity was reduced so that the foot landed in a more plantarflexed position (i.e., with a smaller angle between the foot and floor). Reducing the foot-floor angle has been commonly cited as part of a proactive strategy used when anticipating stepping onto a slippery surface [12, 16, 26, 29]. Landing with a flat foot increases the amount of surface area in contact with the ground, which likely reduces the chances of slipping. In addition, although gait kinetics were not collected, a reduced foot-floor angle has been shown to contribute to a reduction in shear forces, resulting in a lower operational COF, reducing slip potential [26]. During midstance, there was a gradual decrease in ankle plantarflexor (MG) activity, likely contributing to decreased ankle impedance as stance phase progressed. A reduction in ankle dorsiflexor activity was again observed during late swing, in opposition to the increase in plantarflexor activity observed during the same period. This might suggest that when preparing to make contact with the ground, subjects used a strategy to control foot placement and thereafter switched to an impedance strategy to allow the ankle to remain compliant throughout stance phase. These trends were present in both the self-selected and controlled-speed trials, indicating that a reduction in ankle muscle activity occurred on the slippery walkways independent of the reduction in gait speed. This emphasizes the importance of making these changes at the ankle to aid subjects in walking without slipping on the slippery walkways.

Ankle muscle cocontraction has not been previously reported continuously for the entire stride during steady-state walking. A previous study did evaluate cocontraction about heel contact from −20% to 20% stance, but provided only one discrete value over this interval [14]. In agreement with the reduction in muscle activity seen during midstance, there were trends of reduced CCI on the slippery surfaces, further suggesting a decrease in ankle impedance throughout stance. An increase in CCI was observed from late swing to initial contact for the VS walkway. Even though there was increased activity in MG, there was a decrease in activity in TA, meaning that activity of the antagonistic muscle pair did not increase. This points to a potential limitation when using the CCI to estimate the net activation of a muscle since an increase in the CCI can occur even when there is a decrease in the net muscle activity. Hence, although we have included the CCI for comparison with previous studies [14], it is important to recognize its limitations especially with respect to understanding the net muscle activity about a joint and how this activity contributes to impedance regulation.

The lack of ankle muscle cocontraction found in this study suggests a key difference at the ankle for steady-state walking compared to the anticipation of stepping on a single slippery spot. By continuously adapting to the slippery walkways and gaining experience with the low-friction surface, cocontraction at the ankle became less necessary. Thus, activity about the ankle was modulated differently for steady-state walking compared to stepping onto a potentially slippery spot.

In general, the changes in muscle activity at the knee and hip were in direct opposition to the changes in muscle activity at the ankle, where a gradual increase in activity occurred as the walkway became more slippery. This was true for the knee flexor (ST), hip flexor and adductor (AL), and hip abductor (Gmed). The increase in knee and hip muscle activity resulted in increased knee flexion during stance phase and increased hip flexion throughout nearly the entire stride, which likely lowered the body’s center of mass (COM), improving stability. These changes were anticipated based on previous work [26].

The range of hip abduction/adduction achieved during the stride was significantly less on the slippery walkways compared to the NS walkway. A smaller range of hip frontal plane motion would be consistent with increased hip impedance. Three phases of activity in the hip frontal plane occur during stance phase on a NS surface [60]; an initial concentric contraction of the adductors, followed by a burst of activity in the abductors to control further adduction, and, finally, a concentric contraction from the abductors to return to a neutral position. For each of these phases, we observed an increase in muscle activity on the slippery walkways, for both the self-selected and controlled-speed trials. Initially, an increase in adductor activity resulted in a greater degree of adduction. During midstance, an increase in abductor activity resulted in a decrease in the degree of adduction. Finally, stance phase ended with increased adduction due to enhanced adductor activity, ultimately positioning the hip back into the neutral position for that walkway by the end of swing phase. Enhanced muscle activity at precise times likely provided increased control over frontal plane hip motion on the slippery walkways, which would greatly increase trunk stabilization.

An interesting trend was observed in select knee and hip muscles in which there appeared to be a broadening of muscle activity on the slippery walkways. This has been previously observed in other studies that utilized a walkway similar to the very slippery walkway [15, 61]. Martino *et al.* commented that the widening of muscle activity might be how the nervous system "copes" with the unstable conditions. Interestingly, the opposite trend appears for the ankle muscles in which the EMG patterns are less wide on the slippery walkways. To confirm these trends are significant across all subjects, an additional analysis needs to be completed.

Unlike the ankle, the results for some knee and hip muscles showed large differences between the self-selected and controlled-speed trials that may have obscured interesting trends. When gait speed was controlled, there was a gradual increase in VL activity as the walkway became more slippery and increased Gmax activity on the VS walkway. Heightened VL activity in conjunction with the already heightened ST activity could indicate a potential increase in cocontraction about the knee on the slippery walkways. Heightened knee muscle cocontraction has been previously observed during initial contact when subjects anticipated stepping onto a single slippery spot [14]. Increased cocontraction would likely stiffen the knee joint [50], which would contribute to stabilization of the proximal leg and trunk. Similarly, additional hip muscle activity recruited for the VS walkway could indicate further stabilization of the trunk for the most extreme slippery conditions. For the controlled-speed trials, it appears that a gradient of muscle activity was present across the lower limb; muscle activity was gradually reduced in distal musculature, activity was gradually enhanced as the musculature became more proximal, and, finally, the most proximal muscles showed a consistent increase in activity for the VS walkway alone. This gradation of activity was less apparent for the self-selected trials, suggesting a reorganization of muscle activity as subjects walked at faster, more natural speeds.

2.5.2. Implication of Findings

We found that ankle muscle activity was reduced on the slippery walkways for both the self-selected and controlled-speed trials, highlighting the importance of the ankle in aiding subjects to walk without slipping on slippery surfaces. The decrease in activity suggests a decrease in joint impedance. Further work is needed to determine if these changes in joint impedance inferred from EMG are significant, as well as how they influence the mechanical coupling between joints. Both findings could be useful for the control of prosthetic limbs, as more advanced prosthetic limbs are being developed that are capable of multi-joint impedance control [42]. Additionally, a concern of many commercially available prosthetic legs is they often have a stiff ankle-foot complex. Our results suggest this design could be dangerous when used on a slippery surface. In contrast, designs that modulate impedance to be in accordance with the surface conditions could greatly improve subject safety and locomotor independence.

The role of the knee and hip in ambulation on slippery surfaces is less clear due to the differences between the self-selected and controlled-speed trials. Upon controlling for gait speed, knee muscle activity increased in both muscles of the antagonistic muscle pair on the slippery walkways. This could suggest an increase in cocontraction at the knee, resulting in an increase in joint impedance. This might further necessitate the use of a prosthetic limb capable of multijoint impedance control to improve lower limb amputee locomotion on slippery surfaces. Whether a change in knee impedance is occurring needs to be confirmed with future work. It also remains unclear whether this change would be appropriate under natural circumstances (i.e., when subjects walk at their own prescribed pace in real-life situations). In addition, when speed was controlled, there was an increase in activity across all hip muscles, but only on the VS walkway. This could indicate a need to recruit additional proximal muscle activity for extremely slippery conditions, but additional recruitment is less necessary for moderately slippery conditions.

The changes in muscle activity across the lower limb have additional implications beyond potential changes in joint impedance. Independent of gait speed, enhanced muscle activity at the knee and hip resulted in increased knee and hip flexion, and stabilization of the hip in the frontal plane. Although COM was not directly quantified, it can be inferred that the body's COM was lowered through increased knee and hip flexion. A previous study observed that subjects kept their

COM centered over the supporting limb through lateral stabilization of the pelvis when walking on a VS walkway [15]. Although we did not quantify lateral COM displacements, the range of hip abduction/adduction achieved on both slippery walkways was significantly smaller than that on the NS walkway, suggesting a lateral stabilization of the hip, and, thus, enhanced control over the body's COM. It has been previously shown that precise coordination of the lower limb muscle activity reduces COM motion on a slippery surface [62]. While we cannot directly comment on this based on the data acquired in the current study, it is possible that reduction in COM movement is a key component in the strategy used to reduce slip potential.

To walk without slipping on the slippery walkways, the knee and hip could be increasing stabilization of the trunk by lowering and stabilizing the body's COM. In contrast, the ankle is concerned with keeping a wide area of the foot in contact with the ground, through a reduction in ankle muscle activity that begins before contact with the ground and continues throughout stance phase. The ankle has also been shown to respond passively during the recovery response to an unexpected slip [49], so this reduction in activity might also be the ankle preparing to respond appropriately in the event that a slip occurs. A reduction in cocontraction has been linked to decreased reaction times [50], which would be beneficial in the event of a slip. Additionally, a reduction in ankle plantarflexor muscle activity could be beneficial when considering the storage and release of elastic energy in plantarflexor tendons. A reduction in muscle activity likely results in less shortening of the muscle, and, subsequently, less stretching of the tendon. With less stretching, it is likely the tendons are storing less energy for release during the push-off phase of stance. The plantarflexor muscles and tendons typically act like a catapult during normal locomotion [63], but this catapult effect might be less desirable during ambulation on slippery surfaces and thus it is reduced.

2.5.3. Study Limitations

There were some limitations with the experimental protocol, which could influence the presented results. First, footswitches were used to separate the gait trials into individual strides and then into stance and swing phase. The presence of footswitches under the right heel and toe may have affected subjects' ambulation on the walkways, specifically during heel contact and toe off, important phases for slippery gait analysis. Since these footswitches were placed at the beginning of the experiment and worn in the same position for all walkways, their effect on subject ambulation is likely the same across walkways. Thus, any impact on gait dynamics would be the same and would have no impact on the results presented here.

Another limitation is only a small sample of ankle, knee, and hip muscles were recorded over a limited range of walking speeds. Although we describe changes in muscle activity to be indicative of global changes at the ankle, knee, and hip, these changes have only been confirmed for the recorded muscles and do not necessarily extrapolate to other muscles. For instance, no intrinsic foot muscles were recorded and it has been shown that their biomechanical contribution to locomotion may be separable from extrinsic foot muscles, such as those recorded in this study [64]. Their contribution to locomotion on slippery surfaces cannot be inferred from the present data, but might be an interesting avenue to pursue in future work.

We also did not record COM or kinetics, which could contribute to the strategy used to reduce slip potential. Changes in COM and kinetics were observed in a previous study utilizing a VS walking [15]. Although their results are consistent with the inferences made here, being able to directly report these measures would give more weight to these assertions.

Additionally, we observed some degree of adaptation to the slippery walkways, which may have impacted the presented results. Muscles at the ankle, knee, and hip, showed reduced activity when walking on NS2 compared to NS1. In general, this did not impact the observed trends between the NS and slippery walkways. The decrease in muscle activity on NS2 suggests that gait on the NS walkway was altered by walking on the slippery walkways. Prior experience with a low-friction surface has been shown to alter subsequent gait dynamics [29], so this was an anticipated outcome. In the current experiment, two different slippery walkways were used, meaning that subjects could adapt to one slippery walkway before exposure to the second slippery walkway. Prior exposure to a slippery surface could have affected gait dynamics on the next slippery surface. We did not control for the order in which subjects experienced the walkways (i.e., some subjects experienced MS before VS and vice versa) in the statistical analysis, but the presented results were consistent for all subjects regardless of the order in which they completed the walkways. With such a small sample size within each order (five completed MS first and six completed VS first), additional experiments would need to be conducted to flesh out whether prior exposure to one slippery walkway significantly impacts gait on another slippery walkway. Locomotor adaptation is a complex area of research and to properly understand its effect in the context of slippery surfaces, additional experimentation and careful control studies need to be completed, a potentially interesting area for future work.

2.6. Conclusion

To successfully walk without slipping on a slippery surface requires distinct modulation of muscle activity at the ankle, knee, and hip. There was a gradual reduction in ankle muscle activity as the walkway became more slippery that was independent of walking speed. A reduction in ankle muscle activity likely contributes to decreased joint impedance, which would ensure a greater surface area of the foot was in contact with the ground. There were patterns of increasing muscle activity at the knee and hip on the slippery walkways, which resulted in increased knee and hip flexion and stabilization of the hip in the frontal plane. At matched walking speeds, additional muscle activity was recruited at the knee and hip, potentially contributing to increased cocontraction at these joints. This could mean that increased impedance of these joints might also play a role when ambulating on slippery surfaces. For ambulation over any slippery surface, distinct changes at the ankle, knee, and hip are essential to minimize slipping on these surfaces. A potential change in ankle impedance could indicate that designing a lower limb prosthesis with an ankle joint that behaves similarly to those of able-bodied people under these conditions may reduce the likelihood of a lower limb amputee experiencing a slip over a range of slippery surfaces.

3. Altered Neural Control Reduces Shear Forces and Ankle Impedance When Walking on a Slippery Surface

3.1. Abstract

Walking on a slippery surface is difficult and requires systematic changes across the lower limb. One change is a reduction in ankle muscle activity when the foot is in contact with the slippery surface. In this study, we investigated the consequences of reducing ankle muscle activity on slippery surfaces. We explored two hypotheses often associated with changes in the activity of ankle muscles. The first is reduced muscle activation reduces shear forces to remain below the stiction level that would induce a slip. The second is reduced ankle muscle activity reduces ankle impedance, facilitating better contact between the foot and ground, thereby increasing stiction forces and reducing slip potential. To test these hypotheses, we conducted an experiment with unimpaired adults walking across non-slippery and slippery walkways. Set within the walkway was a mechatronic platform with an embedded force plate used to collect shear forces and to estimate the mechanical impedance of the ankle, parameterized by its stiffness, damping, and inertia. We found a significant reduction in the shear forces in accordance with reduced muscle activity in late mid-stance. We found no significant difference in stiffness between the non-slippery and slippery surface. However, the muscle activation changes that contributed to shear force modulation occurred in late mid-stance, where reliable impedance estimates could not be made due to the foot starting to leave the measurement platform. At the points where impedance could be measured, there was a positive correlation between changes in muscle activation and changes in ankle stiffness across the non-slippery and slippery surfaces. This analysis provided indirect estimates that ankle stiffness was likely reduced later in stance phase, beyond where our

current experimental system allowed direct estimates to be made. Together, these results suggest that reduced muscle activity when walking on slippery surfaces serves to reduce shear forces, and possibly also stiffness, during late mid-stance. Both mechanisms would help to minimize the potential for slipping.

3.2. Introduction

Falls are the second leading cause of death resulting from an accidental injury in the United States [2]. It is thus a substantial economic burden on society, with direct costs of $8.6 billion in 2011 [3]. Fear of falling leads to reduced quality of life, especially for populations known to have high prevalence of falls including the elderly [6] and individuals with a lower-limb amputation [5]. Up to half of all fall-related injuries are the result of a slip [9]. Thus, reducing falls in the presence of slippery surfaces will have a significant impact on the health of individual people and on society as a whole.

A few previous studies demonstrated the compensatory strategies used when individuals are confronted with a slippery surface. When people anticipated stepping on a single slippery spot, they increased ankle muscle activity, which is likely to stiffen the joint [14]. During initial exposure to an entirely slippery surface, which resulted in continuous slipping, people increased activity across all lower-limb muscles, potentially using a limb stiffening strategy to negotiate the slippery surface [15]. After adapting to an entirely slippery surface and successfully avoiding slipping across it, we showed there was a proximal-distal gradient in how muscle activity was controlled. Knee and hip muscles show increased activation on slippery surfaces, whereas the activity of ankle muscles was reduced [65]. From these studies it appears ankle muscle activity is modulated differently whether someone is slipping or preparing to slip on slippery surfaces compared to successfully avoiding slipping.

There are at least two potential explanations for how reduced ankle muscle activity could contribute to slip avoidance. The first is to reduce the shear forces generated when walking on the slippery surface. Stiction forces are reduced on a slippery surface and if the shear forces are not reduced accordingly, a slip will occur. The ankle plantarflexor muscles, along with the tendon, act like a catapult propelling the body forward during stance phase [63]. The shear forces can be reduced by reducing the degree of propulsion, which likely requires a reduction in ankle muscle activity. A second explanation is reducing ankle muscle activity may decrease joint impedance thereby facilitating better contact with the ground. Initiating stance phase with a gentler heel-strike has been shown to be an adaptation to slippery surfaces [15], a finding likely facilitated by reduced impedance.

It is common to infer changes in joint impedance from changes in muscle activity based on previous work during postural tasks that demonstrated they are strongly correlated [7]. Walking is a complex dynamic task, which is very different than the postural task previously studied. Before we can assert the mechanical consequences of changes in muscle activity, direct measurements need to be made of joint impedance when walking on a slippery surface. Additionally, by only assessing the effect of muscle activity on joint impedance, previous studies have overlooked the significant role of ankle muscle activity in modulating propulsive forces during walking [66].

In this study, we investigated the consequences of reduced ankle muscle activity during slip avoidance when subjects continuously walked on a slippery surface. We tested two plausible hypotheses. The first was that reduced activation leads to reduced shear forces, and the second was that it leads to reduced ankle impedance. These hypotheses were tested by estimating ground reaction forces and ankle impedance during steady state walking on a non-slippery and a slippery surface and by comparing these results to the changes in muscle activity. Preliminary findings of this study were reported in abstract form [67].

3.3. Methods

3.3.1. Protocol

Fourteen unimpaired subjects (seven males, seven females; 24 ± 4 years, 65 ± 9 kg, 172 ± 11 cm) gave written informed consent to participate in this study, which was approved by the Northwestern University Institutional Review Board. Subjects wore a harness connected to an overhead gantry system to catch them in case of a fall, though no falls occurred.

All subjects completed the same protocol under non-slippery and slippery conditions in one experimental session. For both conditions, subjects walked across a 7-m long walkway made of laminate floors. Approximately half-way down the walkway, subjects stepped onto a force plate embedded within a mechatronic platform. Rouse et al. previously used this platform to estimate ankle impedance during stance phase of gait [45] (Figure 3.1). We attached the laminate floors to the force plate to maintain a consistent and level walkway. To make the walkway non-slippery, subjects wore treaded socks that provided grip for a higher friction surface (coefficient of friction (COF) > 0.40). To make the walkway slippery, subjects wore soft socks that provided no grip (COF = 0.17±0.01). Previous work, including our own [15, 65], demonstrated prior experience on a slippery walkway affects subsequent gait biomechanics on a non-slippery surface. To ensure we captured true baseline non-slippery walking patterns, all subjects completed the protocol first

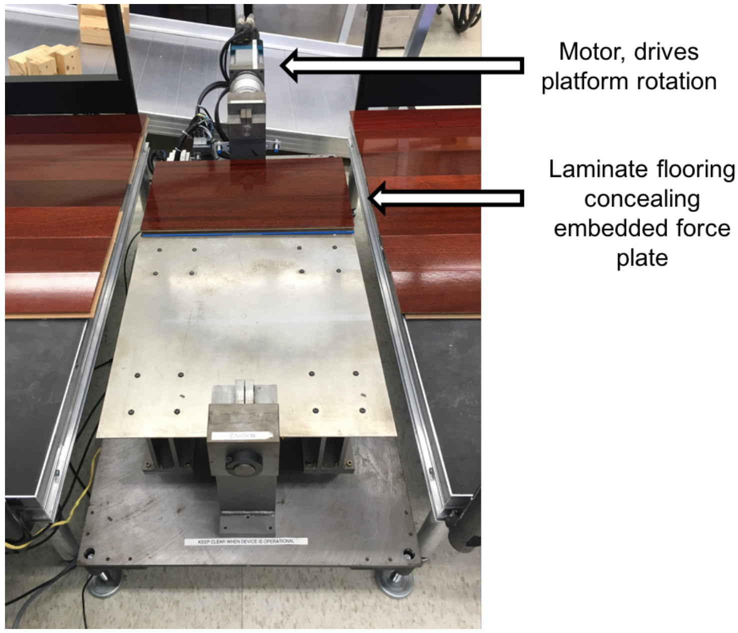


Figure 3.1. Mechatronic platform used to collect ground reaction forces and to estimate ankle impedance. The laminate flooring was placed on top of the force plate to maintain a level walkway. The laminate flooring and concealed force plate compose the part of the platform that rotates, driven by the motor.

under non-slippery conditions, then completed the protocol on the slippery condition.

To maintain a consistent pace between conditions, subjects walked at 85 steps/minute with the aid of a metronome. We chose this pace so subjects could walk comfortably when the walkway was slippery without inducing slips. Subjects completed an initial set of practice trials to ensure accurate pacing. During these trials, the experimenter visually determined the appropriate starting point on the walkway so subjects naturally stepped on the force plate with their right foot and with the center-of-rotation of the ankle aligned to the center-of-rotation of the mechatronic platform. Using high-definition video of the subject's foot and the force plate data, we were able to determine the average misalignment as 0.8±0.9 cm and 1.8±1.2 cm for the non-slippery and slippery conditions, respectively. According to previous work by Rouse et al. that predicts for every cm of misalignment a 5.5% under-estimation of stiffness, this misalignment results in an average under-estimation of 5% and 10% for non-slippery and slippery, respectively [68].

Once subjects consistently stepped with good alignment on the platform at least five consecutive times, the mechatronic platform was used to estimate ankle impedance. This involved applying a rapid perturbation of ankle angle once subjects stepped on the platform, and measuring the corresponding change in ankle angle and ankle torque; these were used to estimate impedance, as described in the Data Analysis section below. These measurements were made while subjects repeated the behavior from the practice trials. Perturbations were applied on random trials with a frequency of 67%, and in a random direction (dorsiflexion or plantarflexion). Each perturbation was a 2 deg "ramp-and-hold." The ramp portion of the perturbation lasted 75 ms and had a constant velocity of approximately 45 deg/second. Perturbations were applied at 170, 320, or 470 ms after heel contact, which was approximately 20, 37, and 55% of stance phase. Both directions were included, even though previous work has shown no effect of perturbation direction on impedance [22], to ensure subjects could not anticipate characteristics of the perturbation.

We chose three time points to capture previously observed changes in muscle activity when walking on a slippery walkway. Based on our previous work, which showed smaller changes in earlier stance and larger changes towards the end of mid-stance, we expected to see the biggest difference in muscle activity at the third time point [65]. Ideally, we would have also included a time point for late mid-stance, but according to previous work by Rouse et al. [22], perturbations occurring later than 55% stance resulted in highly variable estimates of stiffness with a standard deviation of 3.5-4 Nm/rad/kg. We determined this variability was too high to enable detectable changes in stiffness based on the expected changes in EMG. We therefore avoided perturbations beyond 55% of stance. We collected at minimum 20 perturbation trials for each direction and time point. Subjects took a break every 60 trials of walking across the platform and took a larger break between conditions.

We recorded electromyography (EMG) activity using bipolar surface electrodes (model DE2.1; Delsys, Boston, MA, USA) from four ankle muscles: medial and lateral gastrocnemius (MG and LG), soleus (SOL), and tibialis anterior (TA). The signals were amplified 1000x and bandpass filtered with cutoff frequencies of 20 and 450 Hz (Bagnoli 16, Delsys). We measured ankle angle using an electrogoniometer (Delsys, Boston, MA), with one end securely attached to the right shank and the other attached to the inside right foot. We collected ground reaction forces using the force plate (model: 9260AA3, Kistler, Winterthur, Switzerland) embedded within the mechatronic platform. All data were sampled at 1 kHz with a 16-bit data acquisition system (model: USB-6218, National Instruments, Austin, TX) through MATLAB (The Mathworks, Natick, MA). We also recorded high-definition video of each subject's foot placement on the mechatronic platform.

3.3.2. Data Analysis

We performed analyses for the steps occurring on the mechatronic platform, which we isolated using the force data to detect heel-contact and toe-off. Data from the unperturbed trials were used to estimate ground reaction forces and muscle activity, and data from the perturbation trials were used to estimate ankle impedance. To process the EMG signals, we notch filtered to remove 60 Hz noise and full-wave rectified. We normalized the EMG data for each muscle to the maximum value (0.5-s moving average) recorded across all experimental trials.

The methodology for estimating ankle impedance was previously validated and described [22, 45]; it is also briefly recounted here. We low-pass filtered the acquired force and ankle angle data using a bidirectional fourth-order Butterworth filter with a cutoff frequency of 20 Hz. We removed the forces caused by the intrinsic impedance of the mechatronic platform by mapping the acceleration of the robot's motor angle to the forces from the force platform with no subject present and the laminate flooring attached to the platform. We estimated ankle torque by multiplying the projection of the ground reaction force by the distance to the ankle joint [69]. We then isolated the ankle angle and torque response due to the perturbation alone by subtracting the average non-perturbed profiles from the average perturbed profiles. This was done for each perturbation direction and time point. We obtained the impedance parameters by estimating the coefficients of the following second-order equation from the experimentally measured displacements and joint torques.

$$\Delta T = I\Delta\ddot{\theta} + b\Delta\dot{\theta} + k\Delta\theta \tag{1}$$

In this equation, ΔT is the torque response to the perturbation, $\Delta\theta$ is the angular displacement of the ankle caused by the perturbation, and k, b, and I are the impedance parameters of interest: stiffness, damping, and inertia. We used least squares estimation to estimate the parameters over a 100 ms time window starting at the onset of the perturbation. To account for potential covariance between the estimated stiffness and inertia parameters, we determined the average inertia value across time points and conditions for each subject and redid the impedance estimation using a fixed inertia value to estimate stiffness and damping. The fixed inertia value was on average 0.03±0.01 kgm^2. Previous work showed no difference in the impedance parameters between perturbation directions [22], therefore we averaged the impedance parameters across perturbation directions for each time point.

3.3.3. Statistics

We hypothesized that reduced ankle muscle activity during the slippery condition would lead to a concurrent reduction in the shear forces between the foot and the ground and in the stiffness component of impedance. Previous work during a postural task showed increased muscle activation resulted in increased stiffness, while damping was unaffected [8], so we did not expect to see a change in damping. To test for changes in muscle activity and the shear forces throughout stance phase, moving from heel contact to toe off in 10 ms increments, we did a paired t-test using the average value over a 30 ms window on the non-slippery and slippery conditions for each subject. We did this for each muscle and for both the anterior-posterior shear force (Fx) and the medio-lateral shear force (Fy), using Bonferroni correction to account for multiple comparisons. To test for changes in stiffness and damping, we used a linear mixed-effects model where condition (non-slippery, slippery), perturbation time point (early, middle, late), and their interaction were treated as fixed factors; subject was treated as a random factor. We used an F-test to assess the statistical significance of each factor, with significance tested against a p-value of 0.05. If we found significance for any factor, post-hoc testing was completed using Bonferroni correction.

We also wanted to determine if there was a predictable relationship between EMG and stiffness as has been demonstrated previously during postural tasks [7]. To evaluate this relationship, we did an additional analysis correlating changes in EMG (non-slippery minus slippery) to changes in stiffness. To do this, we averaged the EMG over a 30 ms window before the perturbations were triggered and normalized stiffness to the maximum value achieved across conditions to be consistent with the previous EMG normalization. To assess the effect of the time window chosen for EMG quantification, we shifted the center of the analysis window from 20-60 ms before the perturbations were triggered in 10 ms increments, redoing the analysis each time. These times were chosen to account for the electromechanical delay between muscle activation and muscle force, which is known to be quite variable across experimental conditions [70]. We also used the average activity of the three plantarflexor muscles (MG, LG, and SOL). Previous research that demonstrated the positive correlation between EMG and stiffness used a single electrode site to collect the overall triceps surae (MG, LG, and SOL) activity [7]. To give equal weighting to each muscle site and to provide comparable results to the literature, we chose to present the average plantarflexor activity. We used a linear mixed-effects model to relate the change in stiffness between non-slippery and slippery conditions to the corresponding change in EMG. Again, subject was modeled as a random factor. We classified a relationship as significant if the slope had a p-value less than 0.05.

3.4. Results

3.4.1. Ankle Muscle Activity and Shear Forces

We found significant differences in muscle activity for all but SOL during late mid-stance. During late mid-stance, activity was significantly reduced on the slippery condition for MG and TA starting 510 ms after heel contact, approximately 60% stance phase (Figure 3.2 (a) and (d), $p < 0.05$ corrected for multiple comparisons). Thereafter, LG displayed reduced activity on the slippery condition starting 540 ms after heel contact [Figure 3.2 (b)]. The reduction in activity for MG, LG, and TA lasted until 630 ms after heel contact, approximately 70% stance. While SOL

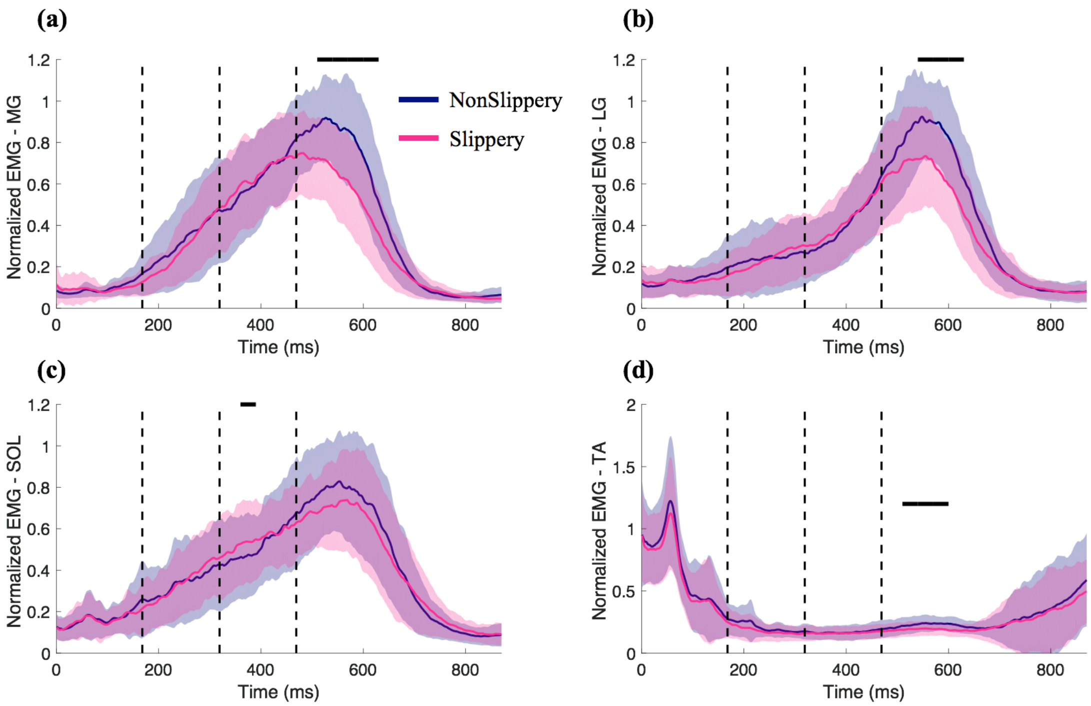


Figure 3.2. Effect of walkway condition on non-perturbed muscle activity. (a) Medial gastrocnemius, (b) lateral gastrocnemius, (c) soleus, and (d) tibialis anterior. Solid blue and pink lines indicate average across subjects (n = 14), shading indicates S.D. Dashed black lines indicate perturbation time points. Horizontal bars indicate periods of significance between conditions ($p < 0.05$).

displayed no differences during late mid-stance, there was a period in which muscle activity was significantly higher on the slippery condition, around 40% stance phase [Figure 3.2 (c)].

We observed a significant reduction in the shear forces both during early stance and late mid-stance. Both the anterior-posterior (Fx) and medio-lateral (Fy) shear forces were reduced on the slippery condition just after heel contact starting at 30 and 60 ms, respectively (Figure 3.3). This initial reduction lasted until approximately 250 ms after heel contact, about 30% stance phase. The anterior-poster shear force was also significantly reduced on the slippery walkway during late mid-stance, ranging from 480-780 ms after heel contact, or 55%-90% stance phase [Figure 3.3 (a)]. During the breaking phase (around 120 ms after heel contact, 15% stance phase), the reduction in Fx on the slippery condition was on average 38%. During the propulsive phase (around 720 ms

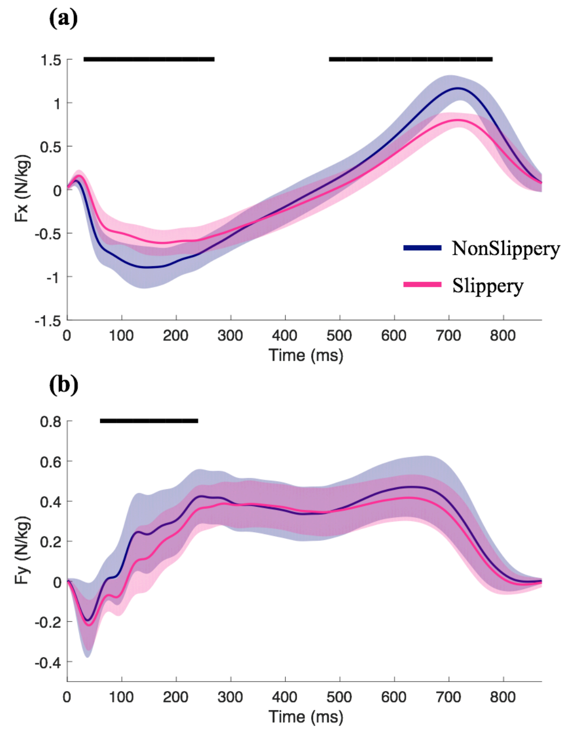


Figure 3.3. Effect of walkway condition on shear forces. (a) Anterior-posterior shear force (Fx) and (b) medio-lateral shear force (Fy). Solid blue and pink lines indicate average across subjects (n = 14), shading indicates S.D. Black lines indicate periods of significance between conditions ($p < 0.05$).

after heel contact, 80% stance phase), the reduction in Fx on the slippery condition was on average 31%.

3.4.2. Ankle Impedance

The second-order parametric fit accurately characterized ankle impedance during this experiment. For a representative subject and perturbation type (Figure 3.4), the torque predicted from Equation (1) agreed well with the actual torque response. Averaged across subjects and perturbation time points, the variance accounted for (VAF) between the actual and predicted torque was 89±5.0% and 89±4.6% for the non-slippery and slippery conditions, respectively (mean ± SD).

The surface on which subjects were walking had no influence on the estimated stiffness

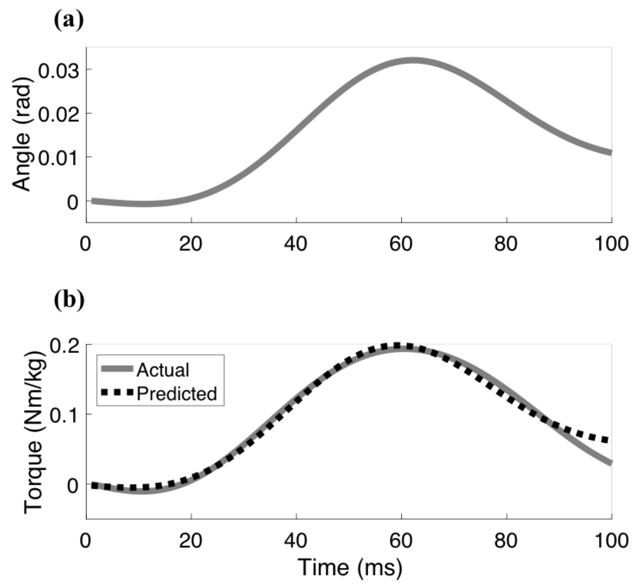


Figure 3.4. Goodness of model fit. Representative perturbed (a) ankle angle and (b) ankle torque for a single subject, with dashed line indicating torque predicted from Equation 1, with inertia fixed.

Table 3.1. Summary statistics for impedance parameters.

	Stiffness		**Damping**	
	FStat	**pValue**	**FStat**	**pValue**
Condition	$F_{1,78} = 0.3$	0.59	$F_{1,78} = 0.05$	0.82
TimePoint	$F_{2,78} = 38.7$	<0.001	$F_{2,78} = 5.9$	0.004
Interaction	$F_{2,78} = 0.9$	0.42	$F_{2,78} = 0.5$	0.61

and damping of the ankle at each of the measured time points (Table 3.1). There was a significant effect of perturbation time point for both parameters, with stiffness increasing and damping decreasing throughout the course of stance phase (Figure 3.5). When we measured impedance, there was no change in muscle activity at or before these times, aside from the small increase in SOL activity on the slippery condition at 40% stance (Figure 3.2, dashed black lines indicate perturbation time points). There was a significant decrease in muscle activity after the perturbations were triggered, when impedance could not be reliably measured. To determine whether we could predict a change in stiffness due to the observed change in muscle activity, it was necessary to establish whether a significant relationship existed between stiffness and EMG.

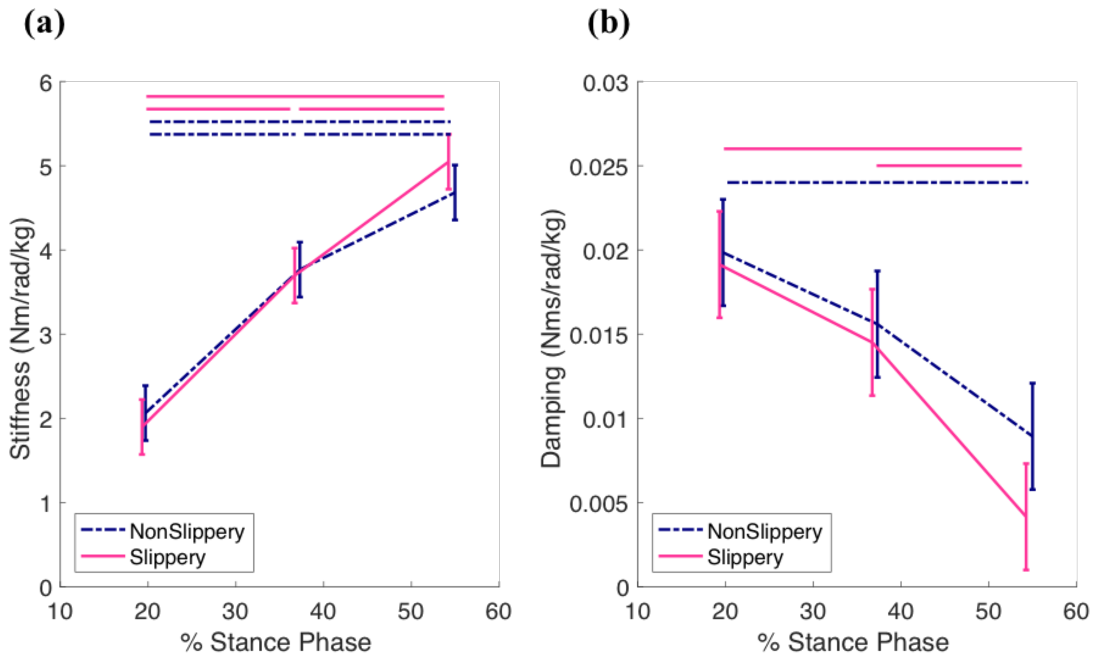


Figure 3.5. Estimated impedance parameters. Parameters averaged across perturbation directions and subjects (n = 14) for (a) stiffness and (b) damping. Error bars indicate S.E. Horizontal bars indicate significantly different time points for each condition ($p < 0.05$).

A change in EMG between the non-slippery and slippery condition was significantly correlated to the change in stiffness. Using EMG quantified 40 ms before the perturbations were triggered, there was a significant correlation between stiffness and EMG (Figure 3.6, p=0.029), such that as the change in EMG between non-slippery and slippery became larger, so did the change in stiffness. This correlation was insensitive to the time window in which EMG was quantified. As the time window moved further from the perturbation time points by 10 ms and 20 ms, the correlation remained significant (p=0.026 and p=0.038, respectively). As the time window moved closer to the perturbation time points by 10 ms and 20 ms, the correlation became gradually less significant (p=0.052 and p=0.075, respectively).

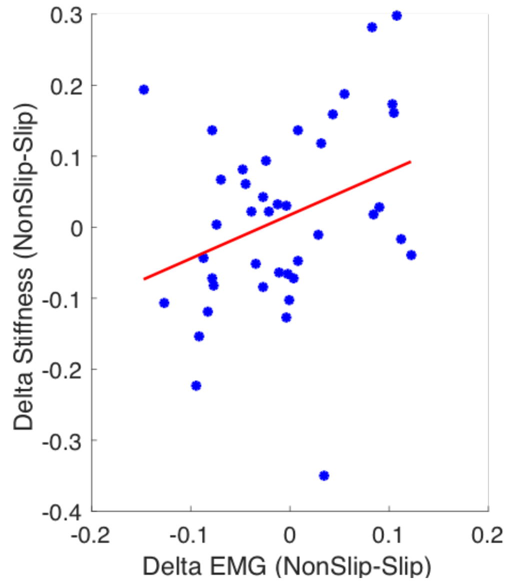


Figure 3.6. Correlation between change in EMG and change in stiffness between conditions. Line estimated using the average slope and offset from a linear mixed-effects model with a random intercept for each subject. The EMG was quantified over a window centered 40 ms before the perturbations were triggered.

3.5. Discussion

In this study, we investigated the consequences of reducing ankle muscle activity when avoiding slips on slippery surfaces. We hypothesized that subjects reduced muscle activity to reduce shear forces and to reduce ankle joint stiffness. We found a significant reduction in the anterior-posterior shear force in accordance with reduced muscle activity in late mid-stance. This indicates while it is necessary to reduce the shear forces to minimize slip potential, the reduction in muscle activity significantly contributes to this change. In contrast, we found no difference in stiffness between the non-slippery and slippery conditions. This is likely because the muscle activation changes that contributed to shear force modulation occurred later in stance, beyond where reliable impedance estimates could be made. Where we were able to measure impedance, we found a significant, positive correlation between changes in muscle activity and changes in stiffness, providing indirect evidence that ankle stiffness was likely reduced later in stance in correspondence with the reduction in muscle activity.

3.5.1. Shear Force Modulation

Our results suggest that reduced muscle activity contributes to reduced shear forces in the anterior-posterior direction, but another way to reduce shear forces during walking is to reduce step length [28]. While we did not measure step length in this study, it is unlikely this had a significant effect on the results. Martin and Marsh previously showed a 10% decrease in step length resulted in an approximately 13% reduction in the anterior-posterior shear force [28]. We see a reduction in Fx of more than 30%, suggesting a change in step length cannot completely explain the decrease in shear forces observed. It is also unlikely subjects reduced their step length by substantially more than 10%. We previously showed when walking on the same non-slippery and slippery surface used in the current study, the self-selected step length was reduced on average by 11% on the slippery walkway [65]. When using a metronome to control the subjects pace, albeit at a slower pace than we used in this study (previous study 75 steps/min, current study 85 steps/min), the difference in stride length between the walkways was reduced to less than 2%. This suggests the difference in stride length in our current study is likely much smaller than 10%, signifying something else must be contributing to the decrease in shear force beyond a potential change in kinematics.

The amount of available friction is reduced on a slippery surface, so to stay above the stiction level and avoid slipping when walking on a slippery surface, the shear forces need to be reduced. The ankle plantarflexors help control the forward momentum of the body during gait [71]. These muscles are reducing activity in preparation for the propulsion phase of stance and subsequently reducing the shear force produced, resulting in sufficient friction to avoid slipping. Modulating the neural drive to the ankle plantarflexor muscles directly contributes to this reduction in slip potential.

3.5.2. Impedance Modulation

The methodology used in this study to estimate ankle impedance provided estimates in agreement with other studies. A previous study analyzing ankle impedance during normal walking from 20-70% stance phase found ankle stiffness varied linearly from 1.5-6.5 Nm/rad/kg [22]. We similarly evaluated impedance from 20-55% stance phase and found stiffness varied on average from 2-5 Nm/rad/kg, falling in line well with Rouse et al. We found damping slightly decreased as stance phase progressed, ranging on average from 0.007-0.02 Nms/rad/kg. These values matched closely to previous work, which for a similar period of stance found damping ranged from 0.01-0.015 Nms/rad/kg [72]. The average inertia value of 0.03 kgm^2 was high when compared to studies evaluating impedance during static conditions, which found ankle inertia to be closer to 0.01 kgm^2 [73]. This is likely due to the small amount of misalignment that occurred between the center-of-rotation of the ankle and the center-of-rotation of the mechatronic platform, resulting in the platform rotation not purely affecting the ankle joint. According to Rouse et al., we can estimate that the 0.8 cm of misalignment for the non-slippery condition and 1.8 cm of misalignment for the slippery condition resulted in an under-estimation of stiffness by 5% and 10%, respectively [68]. The standard error of the stiffness estimates was on average 10% of the stiffness estimates, so it is unlikely a 5% or 10% change in stiffness would significantly affect the results presented here.

While we measured no change in stiffness between the non-slippery and slippery conditions, this is likely because the significant change in muscle activity did not occur until later in stance after the perturbations were triggered. Based on the significant positive correlation between changes in EMG and changes in stiffness, we predict if we perturbed the ankle later in stance, we would observe a reduction in ankle stiffness due to the reduction in muscle activity. Previous work showed if the ankle is perturbed in late stance, when we observed the significant change in muscle activity, estimates of stiffness are highly variable [22] and likely too variable to be able to detect a change in stiffness. Nevertheless, we believe the significant correlation provides evidence that when subjects reduce ankle muscle activity on the slippery walkway, it results in reduced ankle joint stiffness.

The results of this study further solidify our earlier findings on the significance of reducing ankle muscle activity in the presence of slippery surfaces. Previous slipping studies that found changes in muscle activity on a slippery surface assumed subjects adopted a lower-limb stiffening strategy to mirror the increases in muscle activity observed. Specifically, when anticipating stepping onto a single slippery spot, subjects automatically increased ankle muscle and knee muscle co-activation [14] and during initial exposure to a slippery surface, which involved slipping, subjects increased their overall lower-limb muscle activity [15]. In our previous study, we found when the subjects' goal was to *avoid* slipping on a slippery surface, ankle muscle activity was actually reduced [65]. We further confirm those findings here, continuing to show reduced muscle activity on a slippery surface with a mechanical consequence of reduced ankle stiffness. The difference between our results and previous studies highlight how critical it is to appropriately modulate ankle muscle activity for slip avoidance. While the initial reaction when confronted with a slippery surface is to co-contract the ankle muscles, which continues during early exposure to a slippery surface, our results demonstrate that the increase in ankle muscle activation is detrimental for reducing slip potential.

We have also demonstrated that there is a significant relationship between changes in muscle activation and changes in stiffness during walking, which is in agreement with results for static conditions [7]. Other studies that estimated joint impedance during dynamic tasks noted a decoupling between muscle activation and stiffness that was very different than what was observed during static conditions [19-21, 37]. These studies were not designed to assess the relationship between stiffness and muscle activation when other factors known to influence joint impedance, including joint position and joint torque, were controlled. During the stance phase of gait, the ankle joint is increasing towards dorsiflexion, while the level of muscle activation is also increasing, and for this constrained dynamic task, the relationship between stiffness and muscle activation persisted. Though stiffness scaled with muscle activation, the observed relationship was still different than what occurs during static conditions, namely stiffness during walking was much lower than that during static conditions for similar levels of muscle activation. Future work needs to specifically control for the factors that influence joint impedance to assess whether the relationship between muscle activation and stiffness persists for more complex movement.

3.5.3. Limitations

The major limitation of the current study was we could not reliably measure ankle impedance during late mid-stance, where the significant reduction in muscle activity was observed. The Perturberator Robot, the device we used in the current study, was built to measure ankle impedance from 20-70% stance phase, or 10-40% of the gait cycle, with variability in the estimates increasing further into stance [22]. The Anklebot was developed to measure ankle impedance during swing phase and early loading, encompassing 55-100% and 0-5% of the gait cycle [74]. There is a gap in the measurements pooled from these devices from 40-55% of the gait cycle, which is when the change in muscle activity was most important on the slippery walkway. A bridge between these measurements is needed so that future studies under different conditions are not limited by the period of gait in which ankle impedance can be measured.

3.5.4. Implications

The results of this study suggest the neural control employed when walking on a slippery surface plays a critical role in slip avoidance by altering both shear forces and joint impedance. This could have profound implications for the safety of people who have impaired neural control when they come into contact with these hazardous surfaces. Individuals with intact, yet neurologically impaired lower-limbs might have difficultly appropriately modulating their muscle activity, prohibiting them from enacting these critical changes to reduce slip potential. Interventions are being developed to improve paretic propulsion following stroke [75]. While it is important to train for increased propulsion for everyday walking, our results show it is also important to ensure these individuals can modulate propulsion appropriately. Franz and colleagues demonstrated older adults were able to increase propulsion using real-time feedback of ankle muscle activity [66], so a system like this could be adapted to train people to modulate propulsion appropriately for hazardous conditions. Individuals without intact lower-limbs, i.e. amputees, will be entirely reliant on the dynamics of their prosthetic device to enact the appropriate changes. Newly developed powered prostheses can be programmed to provide task-specific mechanical properties [44], as could be useful for walking on slippery surfaces or other terrains. The results of the current study demonstrate how important devices that appropriately modulate their mechanics on different surfaces are for providing safer ambulation for these individuals.

3.6. Conclusion

A key strategy for successfully avoiding slips on slippery surfaces is to reduce ankle muscle activity. We showed reduced muscle activity results in reduced shear forces and we indirectly showed results in reduced joint stiffness. When walking on a slippery surface there are less stiction forces available, so it is imperative to reduce the shear forces produced. The neural control of this movement plays a critical role in reducing the shear forces ultimately leading to slip avoidance. Based on a significant positive correlation between changes in muscle activity and changes in stiffness between the non-slippery and slippery condition, we predict when muscle activity is reduced during late mid-stance, stiffness will also be reduced. These findings have significant implications for individuals who have had neurological injuries as they may be unable to appropriately modulate their level of muscle activity on the slippery surface. These findings also have implications for individuals with a lower-limb amputation because current commercially available prosthetic devices do not adapt the mechanics of the prosthesis for different terrains. Our results indicate this could impair their ability to reduce slip potential on hazardous conditions and also highlight the importance of future research aimed at designing prosthetic devices capable of modulating joint impedance.

4. Ankle Impedance is Decoupled from Joint Torque and Muscle Activation During Dynamic Conditions

4.1. Abstract

The task-dependent adaptation of joint mechanics is important for seamlessly and predictably interacting with our environment. Joint mechanics have traditionally been studied during postural tasks when the factors known to influence joint mechanics can be tightly controlled. Under these conditions, joint stiffness – the static component of joint mechanical properties – is strongly dependent on muscle activation; in the absence of cocontraction, there is a similar dependence on net joint torque. However, the few experimental studies measuring joint stiffness during movement, when joint position and muscle activation continuously vary, showed that these dependencies differ when considering movement rather than posture. Thus, the objective of our work was to determine how joint stiffness is affected by movement and continuously varying muscle activation, when controlling for each factor. We hypothesized that time-dependent changes in muscle activation and joint motion are required to understand how joint stiffness is modulated during dynamic conditions. We tested this hypothesis by determining the independent effect of a change in joint position and change in muscle-generated joint torque on stiffness, as well as the effect of simultaneously varying them. We found that stiffness decreased during movement or during continuously varying joint torque relative to static conditions and the decrease due to varying torque was larger during muscle relaxation. In addition, when joint position and joint torque simultaneously varied, stiffness during eccentric contractions was greater than during concentric contractions. Our results show that to understand how joint mechanics are modulated during functional tasks, which is necessary for helping people for whom joint mechanical

properties have been altered through injury or disease, we must consider the time-dependent changes in joint position and muscle activation.

4.2. Introduction

Our ability to interact effortlessly and predictably with our environment is facilitated by the modulation of the mechanical properties of our limbs. The mechanical properties of a joint can be characterized by impedance, the dynamic relationship between externally imposed changes in joint angle and the torques generated in response. Joint impedance has primarily been studied during posture when joint position and muscle activation can be strictly controlled, due to the increasing complexity in the measurement and interpretation of movement tasks. Knowing how impedance is adapted during movement is important for understanding how the neural mechanisms that modulate impedance are impacted by injury, and for the design of assistive devices that more naturally mimic human behavior.

During postural tasks, there is a predictable relationship between muscle activation, net joint torque, and stiffness, the static component of impedance [7, 8]. Stiffness has been implicated as being significant in the control of posture and movement [32], therefore we focus on its contributions in this work. Based on the results from postural tasks, studies have inferred how stiffness changes during movement using observed changes in muscle activation [14, 15, 17, 18, 76, 77]. However, it is becoming increasingly apparent that there are fundamental differences in the control of limb mechanics during posture and movement [19, 21], and it is unclear how the relationship between muscle activation, joint torque, and stiffness changes during dynamic conditions.

The studies that evaluate the individual effect of changes in joint position or muscle activation provide some clues for how the relationship between stiffness and torque is affected by dynamic conditions. During a transient increase in muscle activation there is a decoupling between stiffness and torque [35], whereas for an imposed movement, stiffness increases following an increase in torque [34]. These studies suggest the relationship between torque and stiffness may be predictable for small imposed movements, but dynamic muscle activation begins to alter this relationship. A few studies on individual muscles have also examined stiffness during dynamic changes in activation or length, but not both. For an isolated muscle-tendon unit, stiffness during shortening is lower than during isometric conditions despite matched forces, while the relationship between stiffness and force during lengthening is similar to isometric conditions [78]. This suggests the muscle length change, which is influenced by the joint position and state of contraction, might impact the relationship between stiffness and torque. For an isolated isometric muscle, the relationship between stiffness and force is dependent on whether the muscle is contracting (increasing force) or relaxing (decreasing force); stiffness changes lead force changes so that at matched forces, a relaxing muscle is stiffer than a contracting muscle [79]. Together, these results suggest that the relationship between joint torque and joint stiffness will depend not only on the magnitude of joint torque, but also the state of the muscles generating that torque.

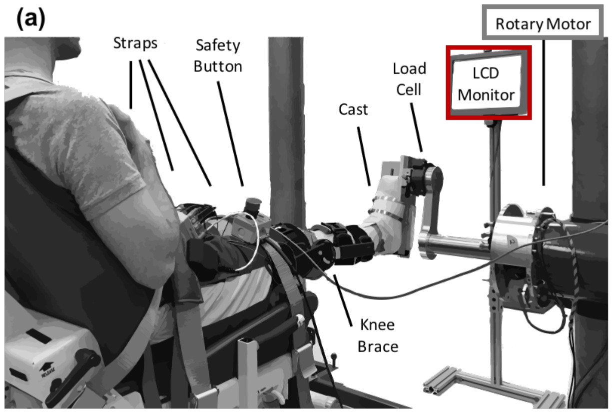
The studies evaluating the effect of simultaneously changing joint position and joint torque are complicated by the fact that the relationship between muscle activation and joint torque can vary dramatically in dynamic conditions due to the force-velocity [80] and other dynamic properties [81-83] of muscle. During voluntary reaching, when joint torque is no longer generated from muscle contraction alone, stiffness does not predictably follow torque, with at times, minimal changes in stiffness despite large changes in muscle activation or torque [19, 20]. The relationships between muscle activation and joint torque, muscle activation and stiffness, and joint torque and stiffness have not been individually teased apart to understand how these three entities are related under dynamic conditions.

The purpose of this study was to investigate how joint stiffness is modulated by movement and continuously varying muscle activation, and to determine how that modulation differs from postural conditions in which joint position and muscle activation are fixed. We hypothesized that the time-dependent changes in joint motion and muscle activation are required to understand how joint stiffness is modulated during dynamic conditions. We tested this hypothesis by determining the independent effect of a change in joint position and change in muscle-generated joint torque on stiffness, as well as the effect of simultaneously varying them. The results from this study help explain why the relationship between joint stiffness, joint torque, and muscle activation is not fixed during movement and signify caution should be used when inferring changes in stiffness based on muscle activation or torque production alone.

4.3. Methods

4.3.1. Subject and Experiment Setup

Ten unimpaired adults (two female, eight male; 27 ± 3 years, 72 ± 9 kg, 177 ± 9 cm) participated in this study, which was approved by the Northwestern University Institutional Review Board. We secured each subject's right ankle to an electric rotary motor (BSM90N-3150AF, Baldor, Fort Smith, AR) via a custom made fiberglass cast [Figure 4.1 (a)]. The cast encased the entire foot, but did not cover the ankle joint, preserving full range-of-motion. We



(b)	Static Position	Dynamic Position
Static Torque	**Block 1:** Static Torque + Static Position (T_SP_S)	**Block 2:** Static Torque + Dynamic Position ($T_SP_{D\uparrow}$ | $T_SP_{D\downarrow}$)
Dynamic Torque	**Block 3:** Dynamic Torque + Static Position ($T_{D\uparrow}P_S$ | $T_{D\downarrow}P_S$)	**Blocks 4 and 5:** Dynamic Torque + Dynamic Position **B4:** $T_{D\uparrow}P_{D\downarrow}$ | $T_{D\downarrow}P_{D\uparrow}$ **B5:** $T_{D\uparrow}P_{D\uparrow}$ | $T_{D\downarrow}P_{D\downarrow}$

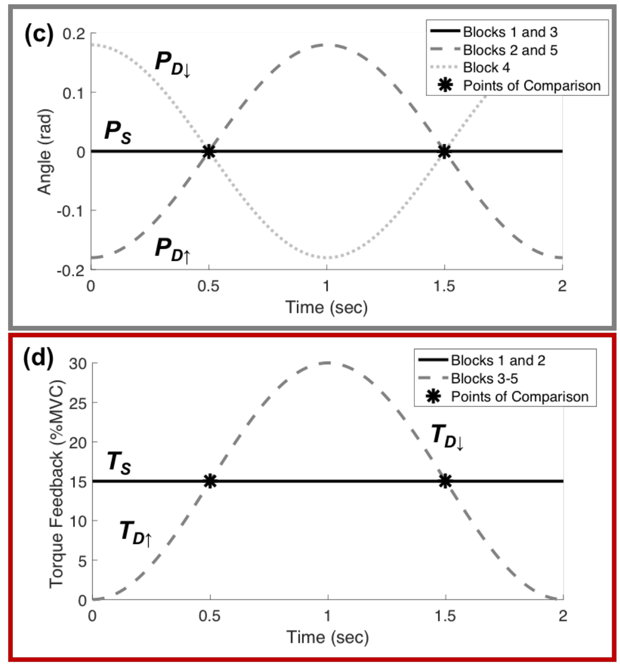


Figure 4.1. Details of the experimental protocol. (a) Subject and experiment setup and (b) blocks of data collection. (c) Position control signals used by the motor, where 0 rad is equivalent to 100 deg between the shank and the foot, positive deflections from 0 rad are increased dorsiflexion and negative deflections are increased plantarflexion. (d) Plantarflexor torque feedback observed by the subject in the LCD monitor.

aligned the ankle to the center of rotation of the motor and restricted movement to the sagittal plane. Subjects sat reclined with their hips at 135 deg and their right leg extended in front of them. We fixed the right knee at 15 deg of flexion using a brace (Innovator DLX, Össur, Reykjavik, Iceland) and secured it, along with the torso, to the chair using straps. We recorded the ankle angle using an encoder integrated with the motor. We used a 6-degree-of-freedom load cell (45E15A4, JR3, Woodland, CA) to acquire force and torque data about the ankle. We controlled the motor using a position control scheme, so the position of the subject's ankle was always dictated by the position of the motor.

We measured the electromyography (EMG) of six ankle muscles— medial and lateral gastrocnemius (MG and LG), soleus (SOL), tibialis anterior (TA), peroneus long (PL), and peroneus brevis (PB)—using bipolar surface electrodes (Noraxon 272, Scottsdale, AZ). We variably amplified the EMG signals (APE-500 & AMT-8, Bortec, Calgary, AB) to maximize the range of the data acquisition system. The analog data were anti-alias filtered at 500 Hz using a 5-pole Bessel filter and sampled at 2.5 kHz (PCI-DAS1602/16, Measurement Computing, Norton, MA). Ankle position was simultaneously recorded using a 24-bit quadrature encoder card (PCI-QUAD04, Measurement Computing, Norton, MA). Data acquisition and motor control were executed using xPC target (The Mathworks Inc., Natick, MA).

4.3.2. Protocol

To isolate the effect of changing position and changing torque on joint mechanics, subjects completed a protocol in which ankle position and joint torque were systematically altered to be static (constant) or dynamic (continuously varying). To fulfill all possible combinations, subjects completed five blocks of data collection [Figure 4.1 (b)]. For all blocks, the motor controlled the subject's ankle position, while subjects were responsible for producing the appropriate level of plantarflexor torque according to feedback shown on an LCD monitor [Figure 4.1 (a)]. We defined an increasing ankle position ($P_D\uparrow$) as moving from plantarflexion to dorsiflexion and a decreasing ankle position ($P_D\downarrow$) as moving from dorsiflexion to plantarflexion [Figure 4.1 (c)]; this notation was chosen to reflect the corresponding length change of the plantarflexor muscle-tendon units resulting from the imposed joint motions. We defined torque as increasing ($T_D\uparrow$) or decreasing ($T_D\downarrow$) if the slope of the plantarflexor torque produced was positive or negative, respectively; again this notation reflected the action of the plantarflexor muscles. The experiment began with collection of maximum voluntary contractions (MVCs) for the plantarflexors and dorsiflexors, with the ankle fixed at the neutral position (Figure 4.1 (c), 0 rad), which was 100 deg measured between the shank and the foot. To produce the position and torque profiles, we used sinusoids as a control signal to the motor and as feedback to the subject [Figure 4.1 (c) and (d)].

Experiments were separated into 5 blocks. Block 1 was the postural condition, in which joint position and torque were static (P_S, T_S). We chose the joint position and torque level to match the state of the ankle during the dynamic conditions (Figure 4.1 (c) and (d), black asterisks). Block 2 evaluated the effect of a dynamic joint position while torque remained static (P_D, T_S). For the dynamic position profile, we used a 0.5 Hz sinusoid with an amplitude of 0.18 rad as the position control signal for the motor (Figure 4.1 (c), dashed gray line). We chose the plantarflexor torque level to match that achieved during the dynamic conditions (Figure 4.1 (d), solid line). Block 3 evaluated the effect of a dynamic joint torque while position remained static (P_S, T_D). For the dynamic torque profile, we used a 0.5 Hz sinusoid with the peak at 30% of the maximum plantarflexor torque achieved during MVC data collection (Figure 4.1 (d), gray dashed line). We used this sinusoid as the feedback signal observed by the subjects in the LCD monitor. We chose the joint position to match that achieved during the dynamic conditions (Figure 4.1 (c), solid line). Blocks 4 and 5 evaluated the effect of a dynamic joint position and dynamic joint torque (P_D, T_D), so we used both the dynamic position and the dynamic torque profiles. For Block 4, the direction of plantarflexor torque production always occurred in the same direction as the effect of the imposed movement, i.e. increasing plantarflexor torque with position decreasing to plantarflexion. This resulted in Block 4 producing a concentric contraction. Due to how we defined the direction of the position and torque, this resulted in position and torque profiles that were out of phase with one another (Figure 4.1 (c) dotted line and Figure 4.1 (d) dashed lined). For Block 5, the direction of plantarflexor torque production always occurred in the opposite direction as the effect of the imposed movement, i.e. increasing plantarflexor torque with position increasing to dorsiflexion. Hence, Block 5 produced an eccentric contraction. This resulted in position and torque profiles that were in-phase with one another (Figure 4.1 (c) and (d), dashed lines).

To estimate impedance, we superimposed a small displacement perturbation on top of each position profile for both static and dynamic conditions. We used a pseudo-random binary sequence (PRBS) with an amplitude of 0.035 rad, velocity of 1.75 rad/sec, and a switching frequency of 0.153 sec.

For the postural condition (Block 1), we collected two 30 sec trials. For the dynamic blocks (Blocks 2-5), we collected 7 trials each lasting 150 sec. The first two trials were practice trials, the first without perturbations superimposed and the second with perturbations. We included practice trials to allow subjects to become proficient at the torque tracking task. Following practice, we collected an additional five trials with perturbations superimposed that were used in the subsequent analysis. The order of the five blocks was randomized for each subject and subjects were allowed to rest for one minute after completion of every trial within a block. We also collected two trials, one using the dynamic position profile and one with the ankle fixed at the neutral position, while the subjects remained completely relaxed.

4.3.3. Data Analysis

Prior to impedance estimation, we completed the following steps to pre-process the data. We decimated the position and torque traces to 100 Hz. The EMG data were notch-filtered to remove 60 Hz noise, full-wave rectified, normalized to the maximum valued achieved during MVC data collection smoothing with a 0.5-sec moving average, and decimated to 100 Hz. The position and torque traces for the dynamic blocks were segmented into three-period long overlapping segments with each segment starting one period after the previous. For the dynamic torque conditions (Figure 4.1 (b), Blocks 3-5), we used the 200 realizations with the lowest mean-squared error relative to the average torque profile. For the static torque and dynamic position condition (Figure 4.1 (b), Block 2), we used the 200 realizations in which the plantarflexor torque produced was most constant, i.e. when the range of values produced was minimized. These trials represented 56% of the collected data. Simulation studies using the multi-segment algorithm demonstrated that 200 realizations was sufficient to produce fits with high variance accounted for (VAF) [84]. Finally, we removed the ensemble means from the position and torque traces.

We estimated ankle impedance using a multi-segment algorithm previously developed in our laboratory [84]. The algorithm computed the time-varying impulse response function (IRF) at each point within a single period for the dynamic blocks and across the entire window of data collected for the purely static condition (Block 1). We estimated ankle stiffness, the static component of impedance, by integrating the impedance IRFs. For the purely static condition (Block 1), this resulted in a single stiffness estimate for each trial, which we averaged together. To obtain confidence intervals for the dynamic stiffness estimates, we used a bootstrapping procedure in which the 200 realizations were randomly sampled with replacement to produce a new ensemble of realizations. We then estimated stiffness using this new ensemble. We repeated the procedure 100 times resulting in a distribution of stiffness estimates for each block and subject. Exemplar results for the bootstrapping procedure for a single subject and block are shown in Figure 4.2. In general, we obtained high variance accounted for (VAF) for the passive data collection and postural condition (Block 1), with lower VAF for the dynamic blocks (Table 4.1).

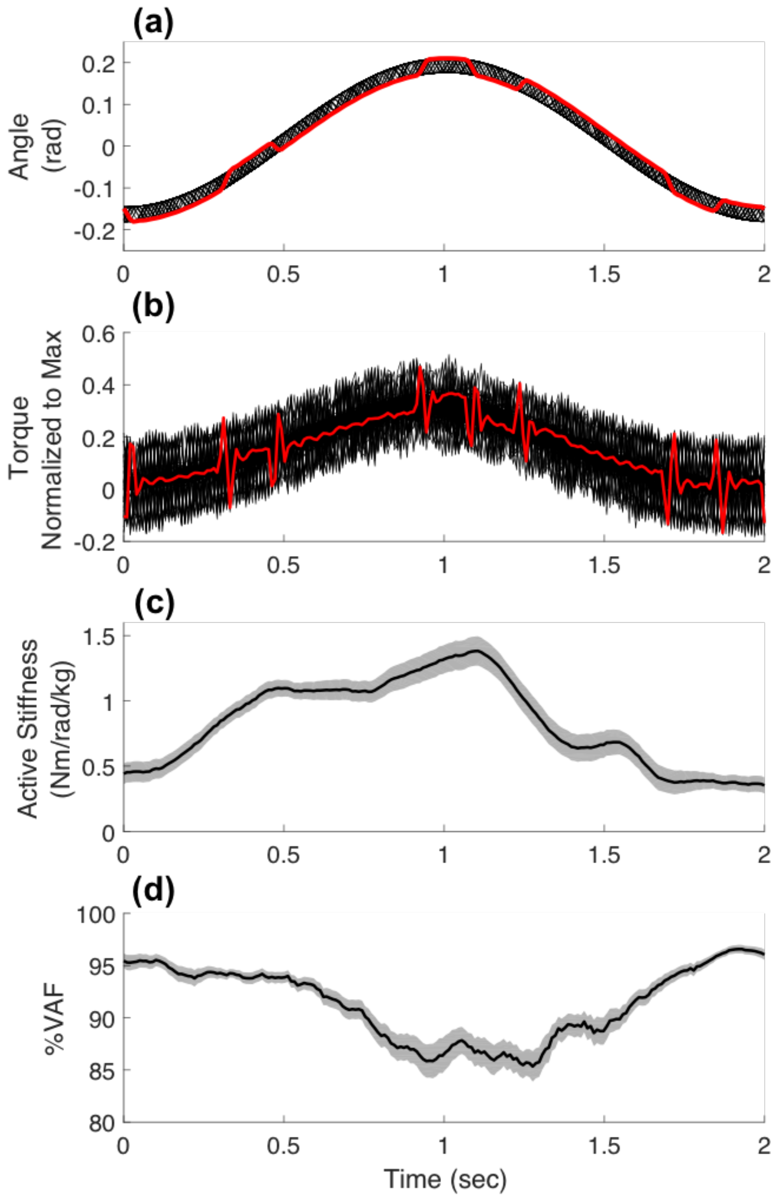


Figure 4.2. Example data for a single subject. All realizations overlaid for (a) ankle position and (b) plantarflexor torque, with red lines indicating a single realization. Average estimated (c) active stiffness and (d) variance accounted for (VAF), with shading indicating S.D. across bootstrap samples. (n = 1)

Table 4.1. Average VAF for each experimental block. (N = 10)

Condition	**Block 1** (T_SP_S)	**Block 2** (T_SP_D)	**Block 3** (T_DP_S)	**Block 4** ($T_{D}\uparrow\downarrow P_{D}\downarrow\uparrow$)	**Block 5** ($T_{D}\uparrow\downarrow P_{D}\uparrow\downarrow$)	**T_PP_S**	**T_PP_D**
%VAF (Mean ± S.D.)	95 ± 3	77 ± 11	83 ± 6	81 ± 7	90 ± 3	98 ± 1	99 ± 1

To standardize the results across subjects, we subtracted the passive stiffness from the estimated stiffness and normalized the remaining active stiffness by subject weight. For the static position conditions (Figure 4.1 (b), Blocks 1 and 3), we removed the passive stiffness obtained at the same static position. For the dynamic position conditions (Figure 4.1 (b), Blocks 2, 4, and 5), we removed the passive stiffness obtained when the subject was relaxed with the continuous imposed movement. We normalized torque to the maximum value obtained during MVC data collection. We organized the EMG into a single plantarflexor group (average activity across MG, LG, SOL, PL, and PB) and a dorsiflexor group (TA).

We also wanted to ensure all subsequent comparisons were done at the same ankle position and plantarflexor torque level. When the ankle position was dynamic (Blocks 2, 4, and 5), we extracted data when the ankle was at the neutral position (Figure 4.1 (c), 0 rad) and averaged across bootstrap samples for each block, resulting in a single data point each for torque, stiffness, plantarflexor EMG, and dorsiflexor EMG when the position was increasing and decreasing. Since Block 3 was collected at the neutral position, we extracted data when the torque level matched the average torque level for the other blocks and averaged across bootstrap samples to have a single data point each for torque, stiffness, plantarflexor EMG, and dorsiflexor EMG when the torque was increasing and decreasing.

4.3.4. Statistics

We hypothesized that time-dependent changes in joint position and joint torque are required to understand how joint stiffness is modulated during dynamic conditions. To test the effect of changing position and torque, we used a linear-mixed effects model with torque direction (static, increasing, decreasing) and position direction (static, increasing, decreasing), as fixed factors with an interaction term and subject as a random factor. This model was used to test significance for torque, stiffness, plantarflexor EMG, and dorsiflexor EMG. We used an F-test to assess the significance of each factor, with significance evaluated against a p-value of 0.05. Specific post-hoc comparisons (Figure 4.3) were completed using Tukey's honestly significant difference to correct for multiple comparisons. We chose the post-hoc comparisons to evaluate the effect of dynamic position (comparing Blocks 1 and 2), the effect of dynamic torque (comparing Blocks 1 and 3), and to evaluate the effect of simultaneous changes in position and torque (comparing Blocks 4 and 5). We completed the data analysis in MATLAB (2016a, MathWorks) and the statistical analysis in R (2016, RStudio, Inc.).

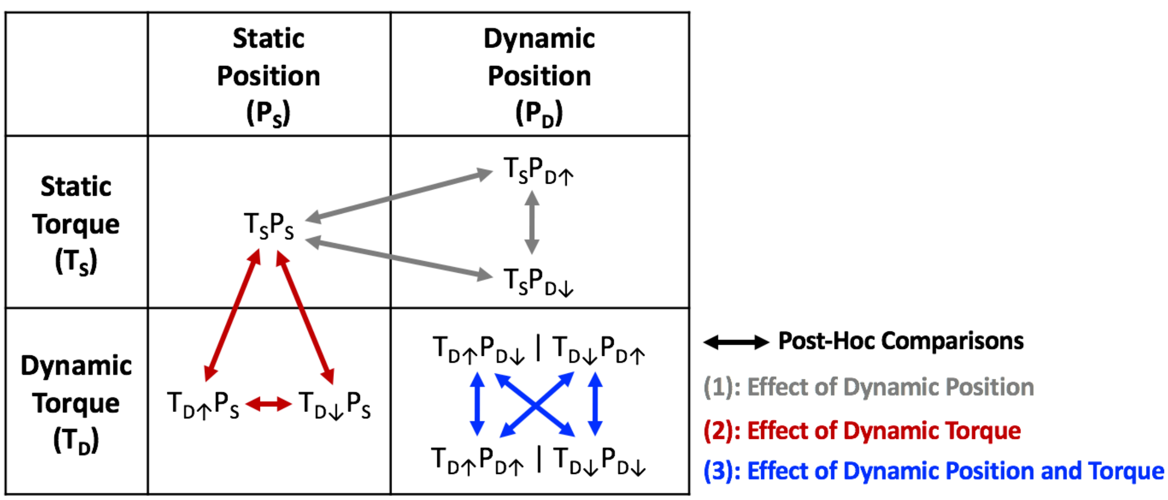


Figure 4.3. Specific post-hoc comparisons completed. (1) Effect of dynamic position, (2) effect of dynamic torque, and (3) effect of dynamic position and dynamic torque. Arrows indicate post-hoc comparisons completed.

4.4. Results

4.4.1. The Effect of Dynamic Position

Ankle stiffness decreased compared to static conditions during a continuously varying ankle position. There was a significant effect of position direction for stiffness ($F_{2,72} = 12.1$, $p <$

0.0001), where stiffness was higher when the position was static compared to dynamic [Figure 4.4 (a)]. When the ankle position was matched (Figure 4.4 (a), black circles), stiffness was significantly higher when the ankle position was static compared to dynamic, regardless of the direction (Figure 4.4 (e); $p < 0.0001$). There was no difference in stiffness whether the ankle position was increasing to dorsiflexion or decreasing to plantarflexion ($p = 0.56$).

Plantarflexor torque was reduced when the ankle position was decreasing to plantarflexion. There was a significant effect of position direction for torque ($F_{2,72} = 12.8$, $p < 0.0001$). The plantarflexor torque achieved when the position was static was not significantly different from when the position was dynamic (Figure 4.4 (b) and (f); $p = 0.37$ increasing position, $p = 0.12$ decreasing position). There was a significant difference in plantarflexor torque between position

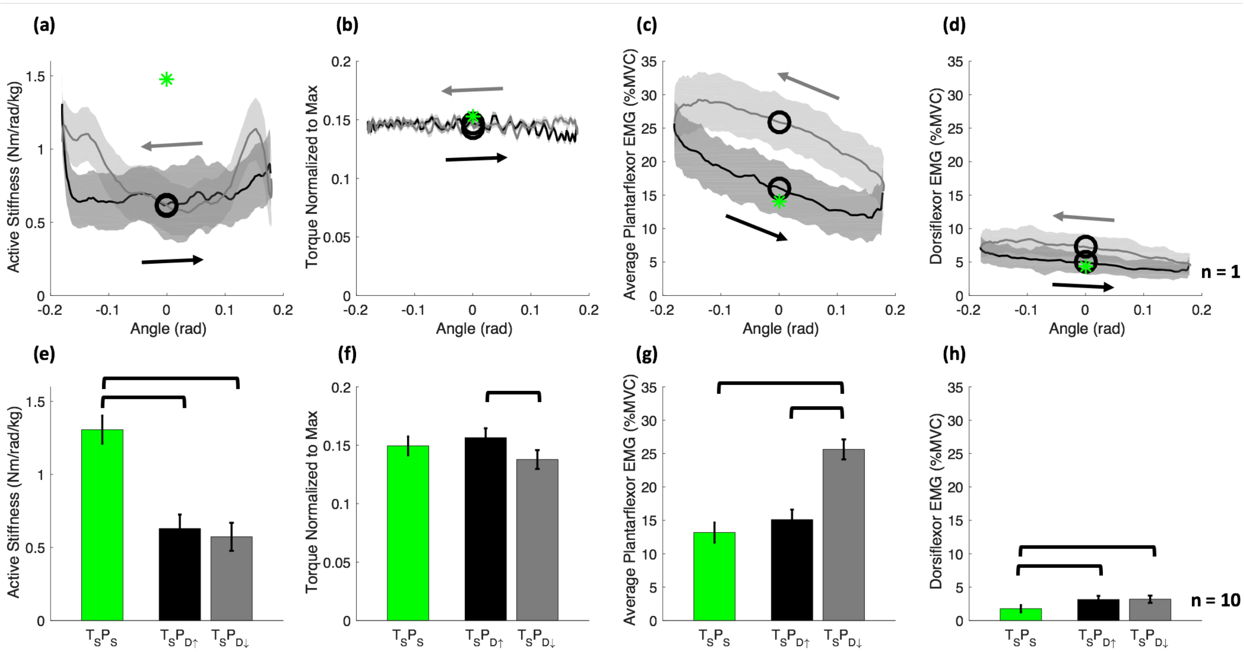


Figure 4.4. Effect of continuously varying joint position during isotonic conditions. Representative data for a single subject (n = 1) for (a) active stiffness, (b) torque, (c) plantarflexor EMG, and (d) dorsiflexor EMG, with shading indicating S.D. The gray lines indicate decreasing position ($P_D\downarrow$) and the black lines indicate increasing position ($P_D\uparrow$), with the black circles at the matched ankle position used in the analysis. Average results across subjects (n = 10) for (e) active stiffness, (f) torque, (g) plantarflexor EMG, and (h) dorsiflexor EMG, with error bars indicating S.E. Horizontal bars indicate significant differences ($p < 0.05$).

increasing to dorsiflexion and position decreasing to plantarflexion ($p = 0.016$).

The changes in EMG were different from the changes in stiffness during a continuously varying ankle position. There was a significant effect of position direction for both plantarflexor EMG ($F_{2,72} = 87.3$, $p < 0.0001$) and dorsiflexor EMG ($F_{2,72} = 5.3$, $p = 0.007$). Substantially more plantarflexor muscle activity was required to maintain a constant torque when the position was decreasing to plantarflexion (Figure 4.4 (c), gray line). When the ankle position was matched (Figure 4.4 (c), black circles), plantarflexor EMG was significantly higher when the position was decreasing to plantarflexion compared to increasing to dorsiflexion or static (Figure 4.4 (g); $p < 0.0001$). The level of dorsiflexor EMG was low when compared to the plantarflexor EMG [Figure 4.4 (c) and (d)]. Dorsiflexor EMG was significantly higher when the position was dynamic compared to static (Figure 4.4 (h); $p = 0.01$ both comparisons). There was no different in dorsiflexor EMG between position directions ($p = 0.94$).

4.4.2. The Effect of Dynamic Torque

Ankle stiffness decreased compared to static conditions during a continuously varying plantarflexor torque. These decreases were greatest during muscle relaxation. There was a significant effect of torque direction for stiffness ($F_{2,72} = 66.5$, $p < 0.0001$). Stiffness was highest for the static condition (Figure 4.5 (a), green star) and there was a large difference in stiffness between plantarflexor torque increasing or plantarflexor torque decreasing (Figure 4.5 (a), red and dark red lines). There was also a significant effect of torque direction for the plantarflexor torque produced ($F_{2,72} = 39.6$, $p < 0.0001$), but there were no significant differences in the torque levels when the ankle position was static (Figure 4.5 (a), black circles; $p > 0.23$ all comparisons). At these matched torque levels, stiffness was highest for static plantarflexor torque production, lowest

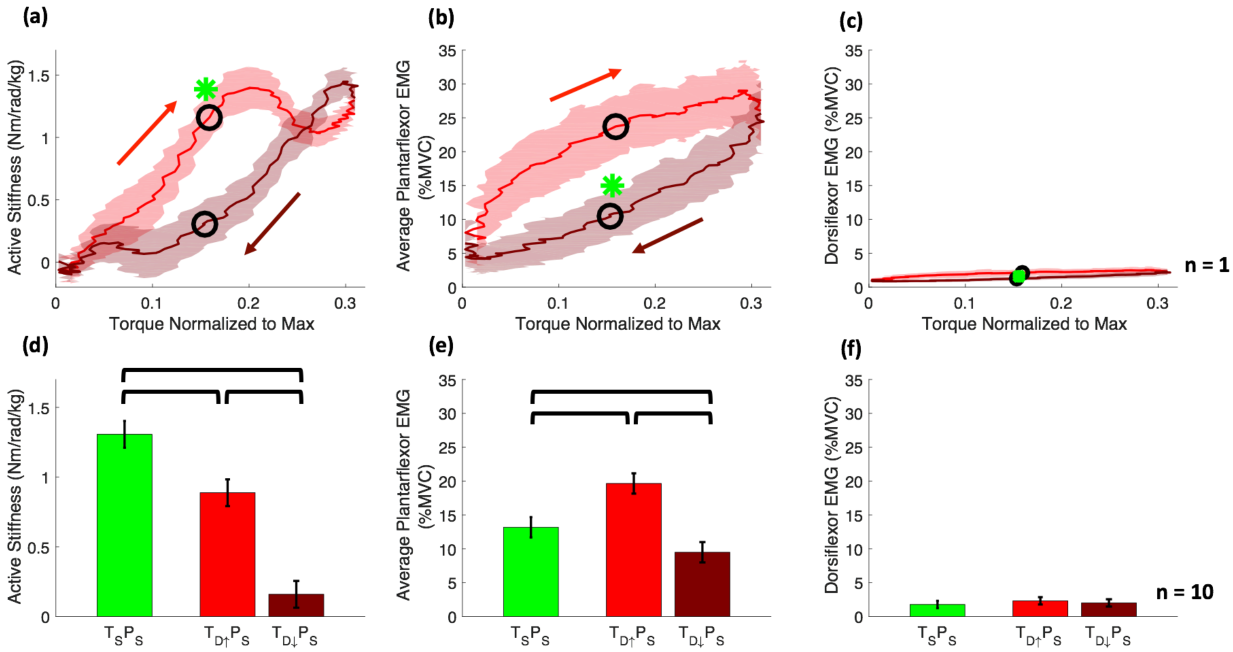


Figure 4.5. Effect of continuously varying joint torque during isometric conditions. Representative data for a single subject (n = 1) for (a) active stiffness, (b) plantarflexor EMG, and (c) dorsiflexor EMG, with shading indicating S.D. The red lines indicate torque increasing ($T_D\uparrow$) and the dark red lines indicate torque decreasing ($T_D\downarrow$), with the black circles at the matched torque levels used in the analysis. Average results across subjects (n = 10) for (d) active stiffness, (e) plantarflexor EMG, and (f) dorsiflexor EMG, with error bars indicating S.E. Horizontal bars indicate significant differences ($p < 0.05$).

for decreasing torque, with increasing torque falling in between (Figure 4.5 (d); $p < 0.0001$ all comparisons).

Plantarflexor EMG was significantly different depending on the torque direction, while dorsiflexor EMG was not. There was a significant effect of torque direction for the plantarflexor EMG ($F_{2,72} = 104.5$, $p < 0.0001$), while there was no effect of torque direction for dorsiflexor EMG ($F_{2,72} = 2.7$, $p = 0.07$). The level of plantarflexor EMG required to achieve the same torque was higher if the muscle was contracting compared to relaxing (Figure 4.5 (b), red vs. dark red lines), with that for the static condition falling in between (Figure 4.5 (b), green star). When torque was matched (Figure 4.5 (b), black circles), plantarflexor EMG was significantly higher for contracting muscle compared to relaxing muscle and plantarflexor EMG for the static condition fell in between (Figure 4.5 (e); $p < 0.02$ all comparisons).

4.4.3. The Effect of Dynamic Position and Dynamic Torque

The plantarflexor torque produced at the matched ankle position was reduced when torque was decreasing during a continuously varying ankle position. There was a significant interaction between position direction and torque direction for the torque produced at the matched ankle position ($F_{4,72} = 11.6$, $p < 0.0001$). There was no difference in torque at the matched ankle position between position directions when torque was increasing (Figure 4.6 left column, black circles; $p = 0.8$). There was a significant difference in torque at the matched ankle position between position directions when torque was decreasing (Figure 4.6 middle column, black circles; $p = 0.03$). Torque was also significantly lower at the matched ankle position when torque was decreasing for both position directions (Figure 4.7 left and middle columns, black circles; $p < 0.0001$ both directions). The fact that torque was not matched across conditions might have some effect on the comparisons for stiffness and EMG, which we keep in mind below.

Ankle stiffness depended on the direction of position and torque changes when both varied simultaneously. There was a significant effect of the interaction between position direction and torque direction for stiffness ($F_{4,72} = 29.2$, $p < 0.0001$). Below, we separate out the effects of the position direction and torque direction for the fully dynamic conditions.

Ankle stiffness depended on the position direction when torque was continuously varying. When plantarflexor torque was increasing [Figure 4.6 (a)] or decreasing [Figure 4.6 (b)], stiffness was different depending on the direction of the position. When plantarflexor torque was increasing, at a matched position and torque level (Figure 4.6 (a), black circles), stiffness was significantly

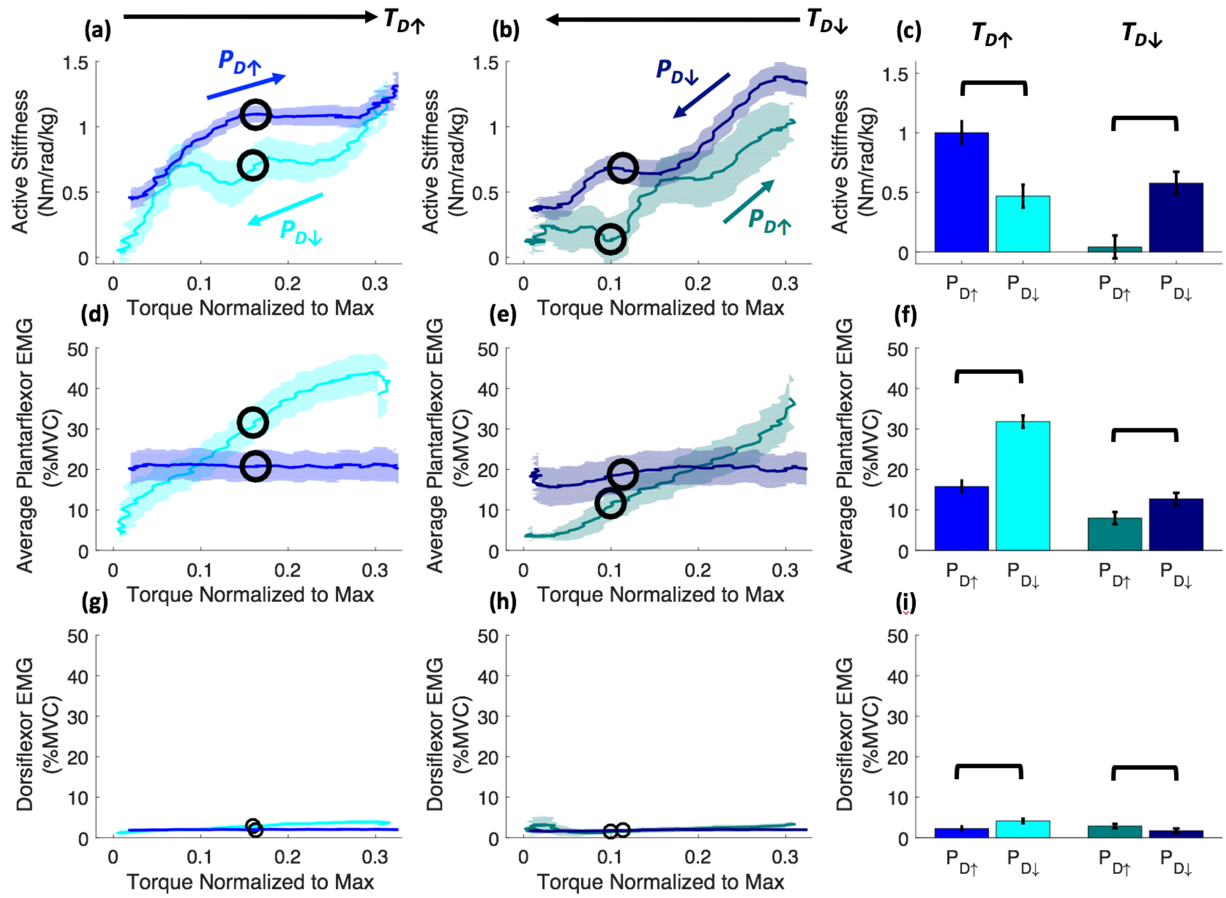


Figure 4.6. Effect of position direction during continuously varying torque. Representative data for a single subject (n = 1) when torque was increasing ($T_{D}\uparrow$) for (a) active stiffness, (d) plantarflexor EMG, and (g) dorsiflexor EMG. Representative data for a single subject (n = 1) when torque was decreasing ($T_{D}\downarrow$) for (b) active stiffness, (e) plantarflexor EMG, and (h) dorsiflexor EMG. Shading indicates S.D. and black circles indicate when ankle position was matched. Group results (n = 10) for (c) active stiffness, (f) plantarflexor EMG, and (i) dorsiflexor EMG, with error bars indicating S.E. Horizontal bars indicate significant differences ($p < 0.05$).

higher for the eccentric contraction (position increasing to dorsiflexion) compared to the concentric contraction (position decreasing to plantarflexion) (Figure 4.6 (c), left bars; $p < 0.0001$). The opposite occurred when plantarflexor torque was decreasing (Figure 4.6 (c), right bars; $p < 0.0001$). For this comparison, torque was not perfectly matched (Figure 4.6 (b), black circles), but the trends between stiffness and torque were similar across torque levels.

The changes in EMG tended to directly oppose the changes in stiffness when position and torque were continuously varying. There was a significant effect of the interaction between position direction and torque direction for both plantarflexor EMG ($F_{4,72} = 10.1$, $p < 0.0001$) and dorsiflexor EMG ($F_{4,72} = 4.7$, $p = 0.002$). There was a linear relationship between torque and plantarflexor EMG during the concentric contraction (Figure 4.6 (d), cyan line), whereas there was no appreciable change in plantarflexor EMG despite a large change in torque for the eccentric contraction (Figure 4.6 (d), blue line). When plantarflexor torque was increasing, plantarflexor and dorsiflexor EMG were higher for the concentric contraction compared to the eccentric contraction (Figure 4.6 (f) and (i), left bars; $p < 0.001$ both comparisons). When plantarflexor torque was decreasing, plantarflexor EMG was higher for position decreasing to plantarflexion (Figure 4.6 (f), right bars; $p = 0.0023$), whereas dorsiflexor EMG was higher for position increasing to dorsiflexion (Figure 4.6 (i), right bars; $p = 0.038$).

Ankle stiffness depended on the torque direction when position was increasing to dorsiflexion, but not when position was decreasing to plantarflexion. For both comparisons, torque was different at the matched ankle position (Figure 4.7 (a) and (b), black circles). The trends between stiffness and torque were consistent throughout the entire waveform, so this is unlikely to impact the results. There was a large difference in stiffness between torque directions when the position was increasing to dorsiflexion (Figure 4.7 (a) and (c), left bars; $p < 0.0001$). In contrast, there was no difference in stiffness between torque directions when the position was decreasing to plantarflexion (Figure 4.7 (b) and (c), left bars; $p = 0.26$).

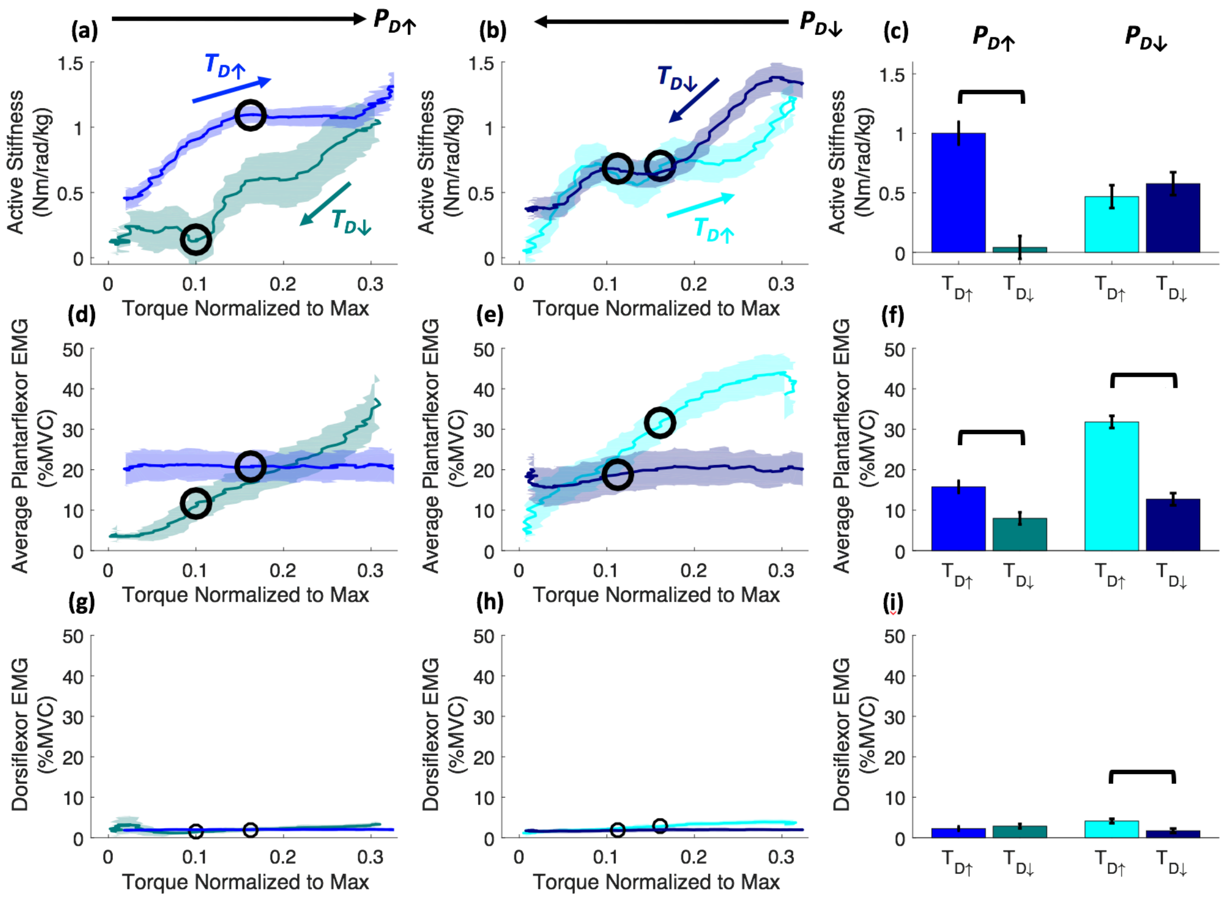


Figure 4.7. Effect of torque direction during continuously varying position. Representative data for a single subject (n = 1) when position was increasing ($P_{D\uparrow}$) for (a) active stiffness, (d) plantarflexor EMG, and (g) dorsiflexor EMG. Representative data for a single subject (n = 1) when position was decreasing ($P_{D\downarrow}$) for (b) active stiffness, (e) plantarflexor EMG, and (h) dorsiflexor EMG. Shading indicates S.D. and black circles indicate when ankle position was matched. Group results (n = 10) for (c) active stiffness, (f) plantarflexor EMG, and (i) dorsiflexor EMG, with error bars indicating S.E. Horizontal bars indicate significant differences ($p < 0.05$).

The changes in EMG did not follow the changes in stiffness when torque and position were continuously varying. There was a dramatic change in stiffness between plantarflexor torque directions when position was increasing to dorsiflexion (Figure 4.7 (c), left bars), while there was only a modest change in plantarflexor EMG (Figure 4.7 (f), left bars; $p < 0.0001$) and no change in dorsiflexor EMG (Figure 4.7 (i), left bars; $p = 0.27$). The torque levels were different at the matched ankle position for these comparisons (Figure 4.7 (d) and (g), black circles). If the torque levels were better matched, it is likely we would see an even smaller difference in plantarflexor EMG, if one at all. In contrast, there was no difference in stiffness between torque directions when the position was decreasing to plantarflexion (Figure 4.7 (c), right bars), but there was a substantial increase in plantarflexor EMG (Figure 4.7 (f), right bars; $p < 0.0001$) and dorsiflexor EMG (Fig 7 (i), right bars; $p < 0.0001$) when torque was increasing. Once again, torque was different at the matched neutral position for these comparisons (Figure 4.7 (e) and (h), black circles). The trends between stiffness and torque were consistent for torque levels higher than 0.1, so it is unlikely the EMG results would be affected.

4.5. Discussion

In this study, we aimed to understand how joint stiffness is affected by movement and continuously varying muscle activation, when controlling for each factor. We hypothesized that time-dependent changes in muscle activation and joint motion are required to understand how joint stiffness is modulated during dynamic conditions. Our results showed that the relationship between stiffness, joint torque, and muscle activation known for postural tasks, does not hold under these dynamic conditions. We found that ankle stiffness decreased during movement relative to static conditions. We additionally observed that ankle stiffness decreased during continuously varying plantarflexor torque compared to static conditions with the decrease more substantial during muscle relaxation. In addition, when joint position and torque were simultaneously changing, ankle stiffness was higher during eccentric contractions compared to concentric contractions, while plantarflexor muscle activity was higher during concentric contractions compared to eccentric contractions.

4.5.1. Joint Impedance Modulation During Changing Position

Ankle stiffness decreased compared to static conditions during movement under isotonic conditions. A decrease in stiffness as a result of movement was previously demonstrated for the knee during passive conditions [21] and for the elbow and knee during voluntary movement [19, 21]. The voluntary movement conditions were complicated by the continuously changing muscle activation and torque. Controlling for the level of torque in this study, we demonstrated under active conditions that stiffness continues to be reduced during movement, for a matched torque level. While torque was nearly constant during the position change, as was stiffness, muscle activation was continuously modulated. As position changed from dorsiflexion to plantarflexion, causing shortening of the plantarflexor muscle-tendon units, EMG increased to maintain a constant torque level. In contrast, when position changed from plantarflexion to dorsiflexion, causing lengthening of the muscle-tendon units, EMG decreased. The modulation of EMG can be explained by the force-velocity relationship of muscle [80], where during shortening a higher level of muscle activation is required to achieve the same force as is achieved during lengthening. Subjects had to compensate for the shortening of the muscle-tendon units by increasing muscle activation so that a constant torque could be maintained, which also resulted in a relatively steady stiffness across the range of joint positions. Our results indicate it is important to consider the length changes of the muscle-tendon unit during movement and to map muscle activation to muscle force, as in this case, torque seemed to be a better indicator of changes in stiffness. It would be interesting to see how stiffness scaled with different isotonic torque levels during changing position to see whether changes in stiffness, or lack thereof, continue to follow changes in torque.

4.5.2. Joint Impedance Modulation During Changing Torque

Ankle stiffness decreased compared to static conditions when plantarflexor torque continuously changed under isometric conditions. This decrease was higher when plantarflexor torque was decreasing due to muscle relaxation. In general, the increase in torque was met with an increase in plantarflexor muscle activity and stiffness, similar to entirely static conditions [7]. A similar relationship between torque, muscle activity, and stiffness was demonstrated at the knee, also under isometric conditions [36]. However, the relationship we observed during varying torque was different from that during the static condition at the matched torque level. For the static condition, a lower level of muscle activity produced the same torque and a significantly higher stiffness. During decreasing plantarflexor torque, the overall relationship between muscle activation, torque, and stiffness was retained, except a lower level of muscle activation was required to achieve the same torque level, which resulted in reduced stiffness. Under these conditions, the changes in stiffness seemed to better follow the changes in muscle activation, while plantarflexor torque remained constant. This also resulted in a hysteresis loop between stiffness and torque, with stiffness higher for a given torque when the muscle was contracting compared to relaxing. In isolated isometric muscle, a similar hysteresis effect was found, except the relaxing phase showed higher stiffness for a given force than the contracting phase [79]. Since we studied the entire joint system, this difference may be attributed to differences in the muscle-tendon dynamics when muscle is contracting compared to relaxing.

4.5.3. Joint Impedance Modulation During Changing Position and Torque

Ankle stiffness depended on the direction of the movement when torque was also continuously varying. This was different than the isotonic condition where stiffness was the same for both movement directions. When the action of the muscle, either increasing plantarflexor torque or decreasing plantarflexor torque, was opposite to the imposed movement, either increasing to dorsiflexion or decreasing to plantarflexion respectively, stiffness was higher. Specifically, this occurred when subjects increased their plantarflexor torque, their plantarflexor muscles contracted and shortened, while simultaneously the ankle was dorsiflexed, which caused lengthening of the muscle-tendon unit. Similarly, this also occurred when subjects decreased their plantarflexor torque, the plantarflexor muscles relaxed and lengthened, while simultaneously their ankle was plantarflexed, which caused shortening of the muscle-tendon unit. The movement might have assisted the muscle in maintaining a constant length, which could explain how the muscle was not contracting or relaxing, but instead maintaining a constant level of activation despite changing torque. We showed for the purely static case that an isometric muscle produced higher stiffness with moderate levels of muscle activation and these results were similar to that when the action of the muscle and the movement were in opposite directions. In contrast, when the action of the muscle and the imposed movement were in the same direction, i.e. increasing plantarflexor torque while the ankle was plantarflexed, muscle activation increased and stiffness decreased. Functionally, when plantarflexor torque was increasing, dorsiflexing the ankle resulted in an eccentric contraction while plantarflexing the ankle resulted in a concentric contraction. Our results showed an eccentric contraction produced higher stiffness than a concentric contraction, even though a concentric contraction required higher plantarflexor EMG than an eccentric contraction.

When position and torque were both changing, there was no singular relationship between stiffness, torque, and muscle activation. During voluntary movement, both position and torque will continuously change magnitude and direction. The effect of these changing dynamics on stiffness cannot be defined by a singular relationship and may explain the complexity observed during voluntary movement tasks. Specifically, during voluntary reaching, Bennett and colleagues showed abrupt changes between stiffness and net muscle torque [19], and these changes may be caused by stiffness being modulated differently depending on the direction of the position change and torque change.

Ankle stiffness depended on the direction of plantarflexor torque when the imposed movement caused lengthening of the muscle-tendon unit, which was similar to the isometric condition. In contrast, when the imposed movement caused shortening of the muscle-tendon unit, stiffness was not dependent on the torque direction. Stiffness was dramatically reduced during muscle relaxation, similar to the isometric condition, when the imposed movement caused lengthening of the muscle-tendon unit. Interestingly, if torque was well-matched, EMG was essentially unchanged. In this case, with EMG and torque constant across torque directions, neither were a good indicator of the dramatic change in stiffness. In contrast, when the imposed movement caused shortening of the muscle-tendon unit, stiffness was unchanged between torque directions despite a dramatic change in EMG. These results suggest that the lengthening muscle-tendon unit behaves similarly to the isometric muscle, while the shortening muscle-tendon unit behaves differently, which was also demonstrated for the isolated muscle-tendon unit [78]. Importantly, when both position and torque are changing, EMG is never a good indicator of changes in stiffness. The fact that such large changes in stiffness do no occur simultaneously with changes in muscle activation and that large changes in muscle activation do not result in changes in stiffness, may indicate complex muscle-tendon dynamics at play.

4.5.4. Physiological Mechanism

The differences observed between eccentric and concentric contractions may be explained by the interactions between the muscle and the tendon during these dynamic conditions. During increasing plantarflexor torque, for the muscle to contract and produce force, it must be able to pull against the tendon. This is seamless when the ankle position is isometric, because the muscle is able to pull against the tendon to contract and shorten, producing force and ultimately stiffness. We saw this for changing muscle activation under isometric conditions, where in general stiffness increased with torque, which increased with muscle activation.

When the muscle-tendon unit undergoes simultaneous length changes during dynamic muscle activation, the interaction between the muscle and tendon becomes more complicated, likely due to high tendon compliance. During passive rotation of the ankle through the same range-of-motion explored in this study, the muscle only takes up on average 27% of the overall length changed of the muscle-tendon unit [85, 86]. While the length change of the muscle is likely different when the muscle is active, it is clear the tendon must be fairly compliant to take up the majority of the length change. Tendon compliance may also be variable during the conditions explored in this study, with Kubo and colleagues showing that tendon compliance decreased from 0-40% plantarflexor MVC [87], the same level of muscle activity achieved in our study.

During the eccentric contraction when the muscle-tendon unit is lengthening, moving from plantarflexion to dorsiflexion, it is actually beginning this cycle in a shortened state. Being in a shortened state, would contribute to greater sarcomere overlap in the muscle, resulting in a fast rise in force, and thus stiffness, at the onset of contraction [88]. As the muscle is contracting, the tendon is rapidly lengthening, taking up most of the overall length change of the muscle-tendon unit. This provides greater resistance for the muscle to continue to contract against, facilitating an increase in force and stiffness.

Conversely, during the concentric contraction when the muscle-tendon unit is shortening, moving from dorsiflexion to plantarflexion, it begins the cycle in a lengthened state. This results in less sarcomere overlap in the muscle, resulting in a slower rise in force and stiffness [88]. While the muscle is delayed to produce force and stiffness, as the position change causes shortening, the tendon rapidly shortens and the muscle must attempt to contract and increase force against the tendon, which is moving in the same direction as the muscle contraction. Ultimately, the muscle works harder (has increased activation), but is not in a position to facilitate either force or stiffness production. In addition, with the tendon become increasing short, potentially near slack length [89], tendon stiffness might become lower than muscle stiffness and thus set overall joint stiffness independent of the changes in muscle activation.

4.5.5. Limitations

One limitation of this study is that with the data we currently have, we cannot conclude whether the muscle is actually shortening, lengthening, or potentially isometric during the different dynamic conditions. Without direct measurements of the changes in muscle or tendon length, as can be collected using ultrasound, we can only postulate on the role of the muscle-tendon dynamics during these conditions. Knowing the state of the muscle and tendon during the continuously varying joint position and joint torque is likely critical for understanding the complex modulation of joint impedance observed.

4.5.6. Implications

Our work shows that knowing the dynamics of the joint position and joint torque are essential for determining the modulation of joint impedance during functional movement. These results have applications for the design of more naturally behaving prosthetic devices meant to more closely mimic the mechanics of the intact leg [43]. These devices are currently being developed to allow ambulation over different terrains with appropriate impedance-based parameters [42]. To extend this work to a variety of different walking conditions, a basic understanding of how impedance is modulated during these movements is needed, and this study helps to build upon this necessary fundamental knowledge. In addition, individuals who have altered neural control, as a result of stroke or neurological disease, will have an altered ability to modulate their limb impedance for different tasks. Before we can understand how the altered neural control impairs their impedance modulation, and thereby better rehabilitate their impairments, we need to know how impedance is modulated during movement in an unimpaired state, as we have explored in this study.

5. Concluding Remarks

5.1. Summary of Findings

The objective of this work was to study the neural and mechanical adaptations that allow us to traverse a multitude of terrains using walking on slippery surfaces as a paradigm to examine these adaptations. We also investigated how concurrent changes in joint torque and joint position, as occur during walking, affect joint stiffness. These findings are instrumental to understanding the link between neural and mechanical adaptations during complex tasks such as walking. The research in this dissertation:

- demonstrates that unimpaired individuals adopt a proximal-distant gradient in the control of muscle activity when walking without slipping on slippery surfaces, with select knee and hip muscles increasing activity while the ankle gradually reduces muscle activity as the walkway becomes more slippery (Chapter 2);
- determines that the reduction in ankle muscle activity when walking on slippery surfaces serves to reduce shear forces and ankle joint stiffness during late mid-stance, both of which help minimize the potential for slipping (Chapter 3);
- shows that continuous changes in joint position or joint torque reduce ankle stiffness compared to static conditions (Chapter 4)
- shows that the reduction in stiffness due to varying joint torque is larger during muscle relaxation (Chapter 4)
- shows that when simultaneously changing joint position and joint torque, the stiffness during an eccentric contraction is higher than during a concentric

contraction, even though a concentric contraction requires higher muscle activation than an eccentric contraction (Chapter 4).

5.2. Implications

We continuously adapt our gait biomechanics for different terrains during normal walking. These adaptations require systematic changes in lower-limb kinematics, kinetics, and muscle activation. These changes have mechanical consequences, adapting our joint impedance to meet the demands of the different locomotion tasks. In the case of walking on slippery surfaces, the adaptation of joint impedance directly contributes to our ability to avoid slipping, so appropriately modulating joint impedance is crucial to improve our safety when interacting with these conditions. If the mechanisms that modulate joint impedance are injured or altered, through the process of aging, from neurological diseases, musculoskeletal injury, amputation, etc., it might be more difficult or nearly impossible to employ these adaptations, resulting in increased incidence of falls.

Older adults are known to have a high prevalence of falls [4] despite the fact that they naturally adopt a more cautious gait pattern when walking [13]. The major characteristics associated with a cautious gait pattern are reduced stride length and a smaller and more slowly changing foot-floor angle at heel-contact [13, 26]. These modifications alone clearly do not help older adults avoid the initiation of a slip and subsequent fall. We showed in Chapter 2, that increased knee and hip muscle activity and reduced ankle muscle activity are also part of a cautious gait pattern in young adults, but whether older adults also adapt their neural control when they employ a cautious gait pattern is not known. During normal over-ground walking, older adults were shown to increase muscle coactivation at the ankle and the knee compared to young adults [90], and if the increased coactivation at the ankle is not properly modulated when walking on a slippery surface, they may be more inclined to slip and fall. We know that calf muscle-tendon properties are altered during aging [41], which alters overall joint impedance. Older adults may be adapting their neural control appropriately, but it is possible how this neural control translates to joint impedance modulation is different than what we showed in young adults, and prevents them from adapting the same appropriate strategies for walking on a slippery surface. Our work demonstrates the importance of adapting neural control on different terrains for modulating joint impedance, and if we better understood how these factors are affected by aging, we may provide insight into why older adults fall more often and how we can improve their interactions with hazardous conditions.

Individuals with impaired neural control following a stroke or caused by neurological disease may be unable to appropriately adapt their neural control on different terrains. In Chapter 3, we demonstrated that this might impair their ability to appropriately reduce their shear forces and joint impedance on slippery surfaces to minimize slip potential. Following stroke, individuals who suffer locomotor deficits often have difficulty producing the necessary propulsive forces for community ambulation, but recent work suggests that there are treatments that can help improve paretic propulsion [75]. Our work suggests that in addition to training for increased propulsion, it is also important that these individuals are trained to modulate their propulsive forces appropriately depending on the locomotion task. Franz and colleagues showed that older adults were able to increase their propulsive forces using real-time feedback of ankle muscle activity [66]. Use of feedback may also be helpful for training appropriate modulation of muscle activity on hazardous terrains. It would be interesting to see if this training could be applied to post-stroke individuals to encourage modulation of muscle activity and to also assess how changes in muscle activity are translated to changes in joint impedance during these tasks.

Finally, individuals with a lower-limb amputation are dependent on the mechanics of their prosthesis, which currently do not allow for modulation across different terrains. Our work highlights the importance of providing these individuals with prosthetic devices that can appropriately adapt the mechanics of the device on different terrains to fully restore their natural community ambulation. Chapters 2 and 3 demonstrate that this technology might also be critical to give these individuals a better chance for avoiding slips and falls. Another concern is many commercially available prosthetic legs have a stiff ankle-foot complex, which when encountering hazardous conditions, might predispose them to being more likely to initiate a slip and fall. Designs that modulate impedance to be in accordance with the task demands of different terrains are necessary for improving subject safety and independence in the community.

5.3. Limitations and Future Directions

The major limitation of this work is the difficulty of estimating impedance of an isolated joint during functional movement. This limitation is common to numerous studies evaluating joint impedance during functional movement. As we demonstrated in Chapters 2 and 3, changes in muscle activation during movement have mechanical consequences. However, there is not a simple relationship between muscle activation and joint mechanics during dynamic conditions. Instead, the kinematic and kinetic state of the limb must be considered. We explored those relationships in Chapter 4, using more controlled experimental conditions than we observed in walking. Devices like the Perturberator Robot [45] and Anklebot [74] are important for estimating impedance during functional tasks like walking, but we also need the more constrained laboratory experiments for isolating the influence of the individual factors that continuously change throughout movement. By combining these methods, we can see how specific changes in joint dynamics influence impedance and then evaluate how the results translate to different functional tasks. Obviously there is much more that needs to be done to understand how impedance is modulated during movement.

The work in this dissertation focused on how increasing plantarflexor muscle activity or torque affected impedance during dynamic conditions. In Chapters 2 and 3 we evaluated the mechanical consequences of a change in plantaflexor muscle activity during stance phase of gait and in Chapter 4 we evaluated how changing plantarflexor torque during movement affects ankle stiffness. During natural locomotion we do not use the plantarflexors alone, but instead we constantly switch between antagonist muscle pairs, activating the dorsiflexors at heel contact, switching to the plantarflexors during stance phase, and finally switching back to the dorsiflexors for swing phase. Previous work showed that when isometrically switching between the flexors and extensors at the knee, there was a dramatic drop in stiffness at the transition that fell below relaxed levels [36], but why this occurred was not fully understood. During walking and other voluntary movement tasks including reaching, we constantly switch between antagonistic muscle pairs, so better understanding how switching between muscle pairs affects impedance during movement is an important future direction of this work.

In addition, developing a mechanistic framework to explain the modulation of joint impedance observed during dynamic conditions would be an incredibly useful next step for this research. If we can explain how stiffness was altered by changes in joint position and joint torque, we will be better poised to make predictions going forward on how impedance is modulated during functional movement. To develop a mechanistic framework, it is likely we need to know how the muscle and tendon are individually affected by the changes in position and torque. This is potentially most important for exploring the difference between eccentric and concentric contractions observed in Chapter 4. We cannot say with certainty how much of the length changes experienced by the muscle-tendon unit due to the changing joint position and changing muscle activation is taken up by the tendon or the muscle, and this likely impacts the overall stiffness of the joint. Using ultrasound to measure the length changes experienced by the muscle and the tendon during these dynamic conditions can help provide insight on their individual contributions to joint stiffness under these conditions. It can also help determine whether the muscle-tendon dynamics play only a limited role and it is rather other passive structures, such as connective tissue, ligaments, extracellular matrix, etc., that majorly influence the complex impedance modulation observed.

In addition to continuing research on joint impedance during movement, there are several future directions for understanding how we safely interact with slippery surfaces. In Chapter 2, we evaluated the steady state adaptations for walking without slipping on slippery surfaces, which is especially important when we encounter moderately slippery surfaces like smooth hardwood floors. A critical moment when a slip is likely to be initiated is when we first step onto and interact with a slippery surface. Thus, studying the appropriate adaptations during the transition from a non-slippery to a slippery surface is also important. Of particular interest would be to evaluate how the strategy used during the transition changes with repeated exposures. As Marigold and Patla pointed out [16], some of our initial reactions when interacting with slippery surfaces can be updated for following exposures to prevent slipping or improve our reactive strategy in the future. In Chapter 2, we also showed that prior experience with a slippery walkway had some effect on the gait biomechanics employed on a different slippery walkway or a non-slippery walkway. Although we did not have enough subjects to determine whether these effects were systematic across subjects, it would be interesting to see how repeated exposure to a moderately slippery surface affected initial exposure to a very slippery walkway. If prior experience with a slippery walkway actually improves subsequent exposure to a more slippery walkway, this could be an important training paradigm for populations with high incidence of falls. What is clear from previous studies, including our own, is that prior experience with a slippery surface is important for adjusting the automatic strategies employed to be more favorable for interactions in the future.

In addition, while we focused on the proactive strategies employed on slippery surfaces, the data collected for Chapter 2 also has applications for reactive strategies. Specifically, even though we only analyzed steps in which no slipping occurred, throughout the experiment there were steps in which slips were triggered. This provides a dataset in which some steps were successful at avoiding slipping while others were not. McGorry and colleagues previously analyzed the differences in foot kinematics and utilized coefficient of friction between matched slip and non-slip trials produced by the same person walking on the same floor condition [91], finding a significant difference in horizontal heel velocity following heel strike. In this study, they used Teflon to create a moderately slippery spot and did not record muscle activity. Future work for Chapter 2 could similarly analyze the differences in matched slip and non-slip trials across the different slippery walkways to assess if there are patterns of muscle activity or joint kinematics that resulted in the initiation of a slip. This would provide even more information on what parts of the proactive strategy we identified for avoiding slips on slippery surfaces most contribute to slip avoidance. Most importantly, future studies should analyze the gait adaptations employed by high risk populations, including older adults and individuals with neurological impairments, to identify where their strategies differ and how we can better train the use of appropriate strategies.

Finally, this dissertation used walking on slippery surfaces as a paradigm for understanding the neural and mechanical adaptations required for walking on different terrains. This has created a foundation upon which future work can be done to explore the adaptations used on other terrains. A critical component of fall avoidance is the ability to adapt our gait to changing surface conditions and we demonstrated that inherent to this ability is the adaptation of muscle activity and joint impedance. We should continue to explore the adaptations used for other variable terrains commonly found outside of controlled laboratory settings, including compliant surfaces like grass and sand, and uneven terrains including ramps and stairs.

5.4. Lessons on the Scientific Method

Another take away from the research in this dissertation is the importance of not assuming more than can actually be verified from the previous literature. In Chapter 2, upon initial review of the slipping studies that evaluated changes in muscle activity, we hypothesized there would be a global increase in activation across all lower-limb muscles. We based this hypothesis on the general consensus of previous slipping studies that on slippery surfaces we increase muscle activity to increase stability [14-16]. These were the major results that were consistently discussed and concluded. The results that were not discussed, and were often a single line in the results section, were evidence that increased ankle muscle activity was actually detrimental when encountering slippery surfaces [16, 27, 29]. These results mirror our findings in Chapter 2 that while knee and hip muscle activity increased, ankle muscle activity actually decreased during slip avoidance on slippery surfaces. While studies may claim their results and conclusions are facts, it is important to use a healthy amount of skepticism and to verify the results are consistent across studies, and when they are not, to do the analysis yourself as we did in Chapter 2.

We have also repeatedly discussed the fact that numerous studies have assumed changes in joint mechanics based on changes in muscle activity during movement. We also somewhat fall into that group because we hypothesized in Chapter 3 that reduced muscle activity would have a consequence of reduced ankle joint stiffness, although we experimentally tested our hypothesis. The results in Chapter 4, which demonstrate that stiffness is affected by continuously varying joint position and joint torque, further show that extrapolating the mechanical consequences of changes in muscle activation should not be done in the future without fully knowing the kinematic and kinetic state of the limb.

Understanding the limitations of our knowledge is important for producing scientific studies that do not infer conclusions beyond what is known in the community. Our role as scientists is to fill significant gaps in previous work to move our collective scientific knowledge forward. If we conclude more than is actually possible given our current state of knowledge, it becomes difficult to identify the important gaps that need to be filled because they may seem to have been filled by these prior conclusions. We demonstrated in this dissertation it is important to experimentally test any assumptions or conclusions we make, to provide a deeper understanding of our results and to identify the next steps forward to further our knowledge.

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