MODIFIABILITY AND LATERALITY OF WALKING STABILITY IN UNIMPAIRED ADULTS

by

James B. Tracy

A dissertation submitted to the Faculty of the University of Delaware in partial fulfillment of the requirements for the degree of Doctor of Philosophy in Biomechanics and Movement Science

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IN UNIMPAIRED ADULTS

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Approved: ____
John J. Jeka, Ph.D.
Chair of the Department of Kinesiology and Applied Physiology

Approved: ____
Kathleen S. Matt, Ph.D.
Dean of the College of Health Sciences

Approved: ____
Louis F. Rossi, Ph.D.
Vice Provost for Graduate and Professional Education and
Dean of the Graduate College

I certify that I have read this dissertation and that in my opinion it meets the academic and professional standard required by the University as a dissertation for the degree of Doctor of Philosophy.

Signed: ____

Jeremy R. Crenshaw, Ph.D.
Professor in charge of dissertation

I certify that I have read this dissertation and that in my opinion it meets the academic and professional standard required by the University as a dissertation for the degree of Doctor of Philosophy.

Signed: ____

Jocelyn F. Hafer, Ph.D.
Member of dissertation committee

I certify that I have read this dissertation and that in my opinion it meets the academic and professional standard required by the University as a dissertation for the degree of Doctor of Philosophy.

Signed: ____

Thomas A. Buckley, Ed.D.
Member of dissertation committee

I certify that I have read this dissertation and that in my opinion it meets the academic and professional standard required by the University as a dissertation for the degree of Doctor of Philosophy.

Signed: ____

J. Hendrik Reimann, Ph.D.
Member of dissertation committee

I certify that I have read this dissertation and that in my opinion it meets the academic and professional standard required by the University as a dissertation for the degree of Doctor of Philosophy.

Signed: ______________________________

Jessica L. Allen, Ph.D.
Member of dissertation committee

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ABSTRACT

Introduction

In our previous studies of children with cerebral palsy, a population characterized by poor balance and high fall risk, we observed that the children with cerebral palsy did not walk in a manner that was protective against a forward loss of balance (i.e. a proactive strategy) despite exhibiting an impaired response to anterior perturbations (i.e. a reactive response) compared to children with typical development. We also observed an unexpected asymmetry such that the “dominant limb” had more lateral stability in typically developing children. This dissertation aimed to address two previously unexplored relationships to walking stability without the confounding influence of neurological impairment: proactive stability modifications and between-limb laterality across balance domains in unimpaired adults.

Methods

For our first aim, 11 unimpaired adult participants, including six females and five males, completed a study of large anterior and posterior perturbations at multiple walking speeds, and a novel focus on pre-perturbation steps to investigate how stability could be modified proactively under threats to stability. We used the Margin of Stability (MoS), a spatial measure of stability that accounts for the position and velocity of the whole-body center of mass relative to the base of support, as an

indicator of the state of stability should a perturbation occur. For our second aim, 30 unimpaired adult participants, including 15 females and 15 males, completed a study consisting of stability- and mobility-demanding tasks to investigate how between-limb asymmetries in lateral stability during walking were related to between-limb asymmetries across tasks in different balance domains. We used the Inter-Limb Asymmetry (ILA) index to quantify the between-limb asymmetry, which measures the between-limb difference and scales that value to the participant's height.

Results

Aim 1 – With the threat of posterior perturbations, there was an increase in posterior stability at foot strike ($p < 0.001$, mean difference (standard error) 1.70 (0.26) %BH). With the threat of anterior perturbations, there was more anterior instability at mid-swing during stance on the dominant limb compared to no perturbations ($p = 0.005$, 0.63 (0.15) %BH). Proactive modifications to stability were accompanied by shortened step lengths ($p < 0.005$, 1.7 to 2.2 (0.4) cm) and increased step rates ($p < 0.004$, 0.05 to 0.06 (0.01) steps$\cdot$s^{-1}). Specific to walking at the slow speed and with the threat of posterior perturbations, there was also less time spent in double support ($p = 0.042$, 0.99 (0.34) %gait cycle).

Aim 2 – There were no significant correlations between the ILA for minimum lateral MoS during walking and the respective ILA values for each task (Pearson $r < 0.327$, $p > 0.119$). Stepping with the non-preferred limb changed the task mechanics. When initiating gait with the non-preferred limb, the mediolateral displacement of the center of pressure during the anticipatory postural adjustment phase was significantly greater ($p = 0.04$, $d = 0.43$, mean difference (SE) 0.47 (1.09) cm). At foot strike when stepping with the non-preferred limb after simulated trips, participants had a larger anterior distance ($p = 0.04$, $d = 0.40$, 2.07 (5.12) cm) between the center of mass and the anterior edge of the base of support, with the center of mass posterior to the edge of the base of support. At foot strike when stepping with the non-preferred limb after simulated slips, participants had a larger posterior distance ($p = 0.04$, $d = 0.45$, 4.07 (9.14) cm) and a smaller lateral distance ($p = 0.01$, $d = 0.65$, 2.55 (3.90) cm) between the center of mass and the posterior edge of the base of support, with the center of mass anterior and medial to the edge of the base of support.

Discussion

Beneficial modifications to posterior MoS at foot strike are indeed possible in an unimpaired population within a given walking speed. These proactive modifications to stability were implemented despite the capacity for unimpaired participants to rely on their ability to recover from perturbations. Consequently, anteroposterior stability may be a feasible target for fall-prevention interventions by targeting decreased step lengths or increased step rates while maintaining the same walking speed.

Lower-extremity laterality may be specific to movement tasks or balance domains rather than a representation of systemic lower-extremity impedance/predictive control. Participants did exhibit strong tendencies towards limb preferences within tasks, but commonly used self-reported limb dominance did not seem to predict those preferences. Participants also showed between-limb differences when performing stepping tasks with the non-preferred limb.

Conclusions

This work provides a framework with which to interpret walking stability and asymmetries in additional populations with and without impairments. Unperturbed walking stability is a modifiable risk factor of falls during walking. Lower-extremity laterality may be present but was not correlated across balance domains or explained by self-reported limb dominance. Future studies should not assume between-limb symmetry in stability or assume that self-reported limb dominance is a meaningful predictor of task preference or performance.

Chapter 1

INTRODUCTION

1.1 Overview

Falls are the leading cause of unintentional injury and the third leading cause of unintentional injury-related death in the United States, nearly equivalent to motor vehicle deaths [1]. An increased risk of falls and injury, resulting in a loss of mobility or independence, also significantly impacts the quality of life across populations at risk for falling. For example, decreased mobility is the strongest predictor of death in patrons of assisted-care residences [2], and fall risk is a prevalent concern related to loss of social interactions, decreased self-care, and less productivity across several populations with neuromuscular impairment [3–6]. A fall also reduces self-efficacy [7–9] leading to activity avoidance to prevent a future fall [10–13]. In turn, this sedentary lifestyle reduces life expectancy, cardiovascular health, and neuromuscular function [14–16]. Therefore, addressing the risk of falling is an important aspect of preventing injury and sustaining an individual's quality of life.

Falls while walking represent 33-71% of reported falls across many populations, such as stroke survivors, individuals with Parkinson's disease, young, middle-aged, and older adults [17–24]. We propose that, from a biomechanical perspective, the ability of an individual to prevent a fall after a walking perturbation is influenced by two intrinsic factors: (1) the stability of unperturbed walking (i.e. the "initial conditions" before a perturbation), and (2) balance reaction capabilities after a perturbation (i.e. the "recovery skill", Figure 1.1). As a complement to our previous work on balance reactions [25–32], this dissertation focused on the former aspect of gait stability, investigating proactive modifications to walking stability and the influence of lower-limb laterality.

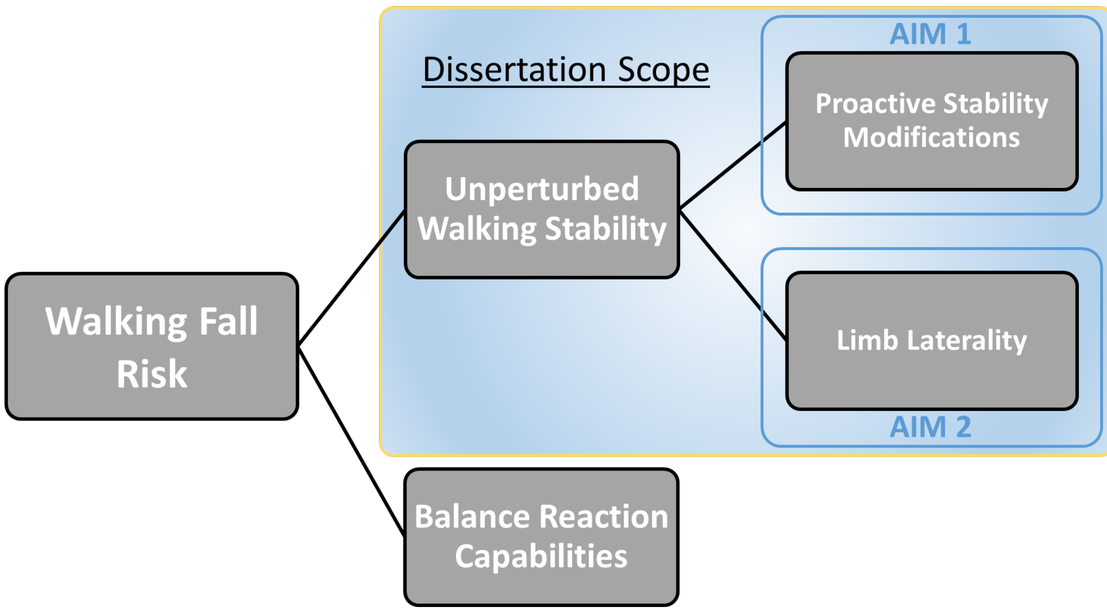

Figure 1.1: **Dissertation scope.** Fall risk during walking is influenced by two major intrinsic factors: the state of stability during walking and a person's recovery capabilities after a disturbance or perturbation. This dissertation focused on two aspects of walking stability: proactive stability modifications and limb laterality. Chapter 2 investigated proactive stability modifications through changes in stability (as measured by the margin of stability) and changes in walking mechanics under threats to stability. Chapter 3 explored relationships between lateral margin of stability asymmetry while walking and limb laterality across different balance domains.

Stability is a term with broad definitions used in previous studies [33]. Within this dissertation, stability was defined as the "capacity of a system to respond to perturbations" [34], and gait stability was specifically defined as "not lead[ing] to falls in spite of perturbations" [33]. Walking stability is related to the size of the perturbation needed to elicit a fall [33,34]. In other words, a "stable" gait is one in which a relatively large perturbation is needed to initiate a loss of balance because the initial conditions prior to a perturbation are controlled (i.e. proactive control), and/or the person has the capacity to regain stability after a perturbation (i.e. reactive control). Given a direct, logical relationship to fall risk, gait stability is a relevant target for interventions to reduce falls in at-risk populations.

Stability during walking has been measured using many different biomechanical variables [33]. We elected to focus on the Margin of Stability (MoS) [35], a spatial measure of stability based on the inverted pendulum model [36] that accounts for the position and velocity of the whole-body center of mass relative to the base of support (Figure 1.2). One advantage of the MoS is that it is proportional to the impulse needed to change stability states [35]; therefore, the MoS has explicit biomechanical meaning and is indicative of the initial stability state should a perturbation occur at a given point in time. The MoS has been applied as a measure of dynamic stability during walking across a wide range of study populations, including young adults [34,37], persons with multiple sclerosis [38], stroke survivors [39], older adults and persons with Parkinson's disease [40], and, in our own study, children with cerebral palsy (CP) [41], showing versatility in its applications as an applied tool. From these studies, we learn several things related to the MoS during walking. The MoS can be manipulated through voluntarily changing step width and step length [34]. The MoS was comparable between treadmill and over ground walking despite differences in step width and step width variability between surfaces [37]. Persons with multiple sclerosis demonstrated a conservative gait strategy with slower walking speeds, shorter stride lengths, and wider steps that result in altered MoS values, and the MoS was correlated with fall history [38]. Unimpaired participants were able to maintain their posterior margin of stability and walking speed by effectively manipulating their step length and frequency compared to post-stroke participants [39]. Persons with Parkinson's disease were less unstable during obstacle crossing by constraining the distance between the center of mass and the base of support compared to healthy older adults [40]. Children with CP showed no difference in anterior stability but utilized a more conservative lateral stability strategy compared to children with typical development [41]. Overall, we acknowledge the versatility of the MoS measure to give insight into the state of stability should a perturbation occur and the measure's important relationship with gait characteristics such as step length, step rate, step width, and walking speed.

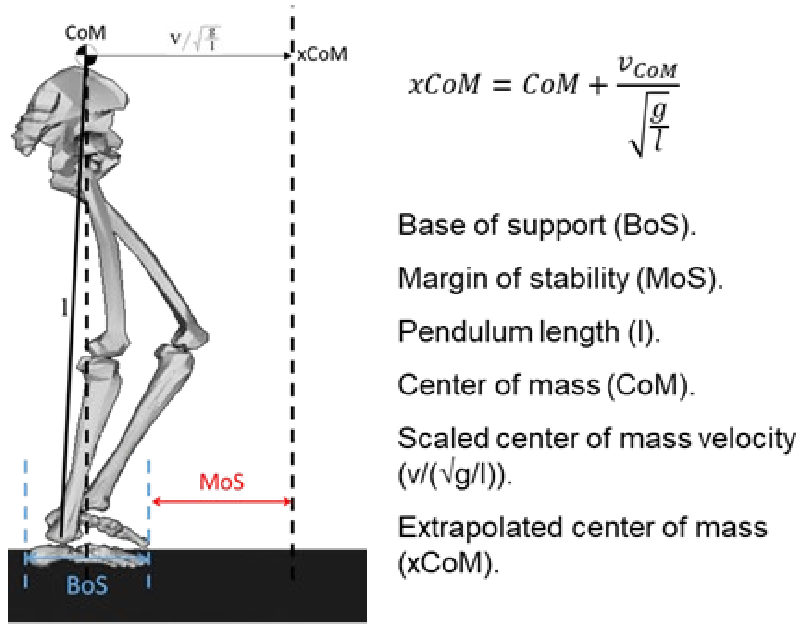

Figure 1.2: **Illustration of the anteroposterior margin of stability at mid-swing.** Figure modified from previous publication [41]. The margin of stability represents the distance between the base of support and the extrapolated center of mass (center of mass position + scaled center of mass velocity). Positive values represent a state of stability (i.e. the extrapolated center of mass is within the base of support or advantageously placed away from the edge of the base of support), and a perturbation is needed to initiate a fall in that direction. Negative values represent a state of instability (i.e. the extrapolated center of mass is outside the base of support), and a compensatory action such as taking a step, applying an external force, or counter-rotating segments about the center of mass is needed to prevent a fall.

The purpose of this dissertation was to address two previously unexplored relationships to walking stability: proactive stability modifications and laterality across balance domains. Incorporating healthy adult participants into these two basic studies allowed us to consider these two aspects of gait stability without the confounding influence of neurological impairment. These studies built upon previously published research by applying two novel testing paradigms. These new approaches brought additional understanding of gait stability at a range of speeds, under multiple conditions, and across balance domains. The knowledge gained from these studies will advance the state of interventions intended to decrease fall risk and to increase regular physical activity participation.

1.2 Significance and Innovation

1.2.1 Modification of Anteroposterior Stability

In our previous study of children with CP, we observed no differences in anterior stability at foot strike (Figure 1.3A and 1.3B) or mid-swing (Figure 1.3C and 1.3D) comparing those with and without CP [41] despite those children with CP exhibiting an impaired response to anterior perturbations [26]. In other words, the children with CP did not walk in a manner that was protective against a forward loss of balance. This result led us to question if, in the absence of neuromuscular impairment and with fully developed gait and balance, anteroposterior stability can be modulated.

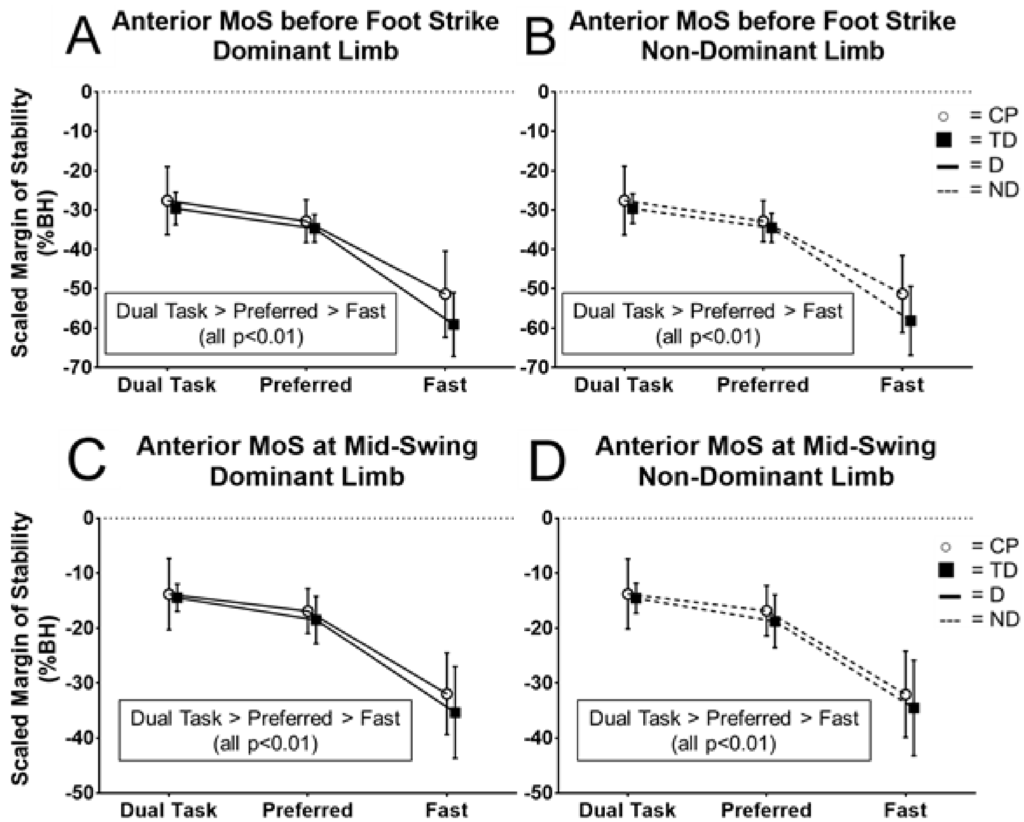

Figure 1.3: **Anterior margin of stability for children with and without cerebral palsy.** Figure modified from previous publication [41]. Anterior margin of stability (MoS) is displayed across different conditions (verbal/mental dual task, preferred walking speed, fast walking speed), between groups (children with cerebral palsy and children with typical development), and between limbs (dominant and non-dominant). Open circles represent the cerebral palsy (CP) group. Closed squares represent the typically developing (TD) group. Solid lines represent the dominant (D) limb. Dashed lines represent the non-dominant (ND) limb. Compared to preferred speeds, both groups were more stable (i.e. less negative MoS) during the dual task condition (p<0.01) and less stable at fast speeds (p<0.01). We observed no between group differences.

Previous studies have investigated the effects of perturbations during walking on stability using a variety of methods. Participants were typically asked to respond to an audible, visual, or verbal prompt to adjust their movement pattern to achieve some movement goal (i.e. modify step width, frequency, or length; perform lateral stepping maneuvers; or reach body positioning targets with knees) [34,42–46]; respond to an overground physical perturbation such as an obstacle inducing a trip [47–50] or a slick surface inducing a slip [47,51,52]; or respond to a physical perturbation such as a waist pull [53–55] or surface translation [43,46,51,52,56–63]. While responding to mediolateral perturbations while walking on a treadmill, participants often adjusted their gait to accommodate the potentially increased fall risk by increasing step width [42,43,53,61], increasing step frequency [53,60,61], and decreasing step length [43,60,61]. These kinematic adjustments were typically accompanied by maintained or increased stability as measured by the MoS [43,53,60,61]. One of these studies also incorporated anteroposterior perturbations, but reported no differences in step width, frequency, or length [60]. Also, the magnitude or intensity of the perturbations that were used is unclear, thus limiting the inferences to this proposed work. Two unknowns remain: the extent to which proactive modifications of stability are utilized and the influence of threats to stability from large anterior or posterior perturbations.

For anteroposterior gait stability to be a target for fall-prevention rehabilitation, it must be modifiable. A previous study with some similarities to our completed protocol did observe modifications in anteroposterior stability of healthy adults related to adaptations of step length, frequency, and width [61]. Therefore, there is some evidence that stability can be modified with threats to stability. However, this study used lateral perturbations, averaged pre- and post-perturbation steps across time, and a single, fixed walking speed. For our first aim presented as Chapter 2, we used a more rigorous study using large anterior and posterior perturbations, an analysis of only the pre-perturbation steps, and multiple walking speeds to investigate how stability can be modified proactively under threats to stability. Increasing gait speed is achieved by increasing step length, step frequency, or both; therefore, at faster walking speeds, it may be more difficult to control stability with these gait parameters because the individual may be closer to their functional limits before walking transitions into running. Therefore, we expected that more pronounced proactive modifications would occur at slower speeds.

Many previous studies have investigated how age and neuromuscular impairment alter balance reactions [28,30,48,49,53,64–74]. The initial conditions that exist prior to a perturbation are under-explored, yet they are an important component to maintaining stability. Previous studies have evaluated changes in gait parameters and dynamic stability during walking with mediolateral perturbations [39,42,43,53,60,61]. Only one of these studies [60] also incorporated anteroposterior perturbations, but the magnitude or intensity of the perturbations that were used is unclear, thus limiting the inferences to our completed work. Therefore, we did not know the influence of threats to stability from anterior or posterior perturbations. We used our computer-controlled treadmill (ActiveStep®, Simbex, Lebanon, NH, USA) to simulate trip- and slip-like perturbations, common causes of falls. Using this treadmill, we could deliver these perturbations across a range of walking speeds with precision and consistency [75]. In our previous study of children with CP, we observed no differences in anterior stability at foot strike (Figure 1.3A and 1.3B) or mid-swing (Figure 1.3C and 1.3D) comparing those with and without CP [41], despite those same children with CP exhibiting an impaired response to anterior perturbations [26]. One explanation for this lack of difference could be the low threat of a perturbation from walking. The large, repeatable anterior and posterior perturbations used in this study provided a significant threat to balance, but not consistent falls into the harness. Large perturbations were used in order to encourage modifications to the initial conditions of stability, if they are possible, in order to prevent falling rather than relying only on reactive balance.

The aim 1 protocol presented in Chapter 2 also built on previous studies by evaluating gait parameters and the MoS through a range of walking speeds utilizing anterior (i.e. trip-like) and posterior (i.e. slip-like) perturbations. While increased stability often can be achieved through decreased walking speed, this decrease in walking speed is counter to rehabilitative goals of improving gait speed. We built on previous studies by determining the influence of gait speed on the proactive ability to modify stability. We also expanded on previous work by evaluating the anterior MoS at mid-swing (i.e. when a trip is likely to occur [76]) in addition to the posterior MoS at foot strike. Because we did not know the extent to which proactive modifications of stability are utilized, we analyzed only the pre-perturbation steps (as done in only one study with below-knee amputees [53]) to focus on the effect of the threat of perturbation on dynamic stability, rather than a mixture of pre- and post-perturbation steps [43,46,54,56,60–62] or only the post-perturbation steps [42,48,49,55,57,77] which evaluate some or all of the perturbation recovery response. Our novel approach specifically targeted the proactive stability modifications that were possible with the threat of a perturbation, also identifying the specific gait modifications that were enacted to achieve such alterations in stability.

For our first aim, presented in Chapter 2, we used a rigorous study of large anterior and posterior perturbations, multiple walking speeds, and a novel focus on pre-perturbation steps to investigate how stability can be modified proactively under threats to stability. By addressing these novel aspects, we took important first steps to determining if anterior and posterior walking stability were modifiable rehabilitative targets.

1.2.2 Limb Dominance and Laterality across Balance Domains

In our previous study of walking in typically developing children [41], we observed an unexpected asymmetry such that the “dominant limb” had more lateral stability in typically developing children (Figure 1.4). Our observed asymmetry in gait stability was unanticipated as the symmetry of gait variables in unimpaired people had previously been assumed across a wide range of analyses [78–84]. In our second aim, presented in Chapter 3, we explored how our observed between-limb difference fit within the theoretical construct of laterality.

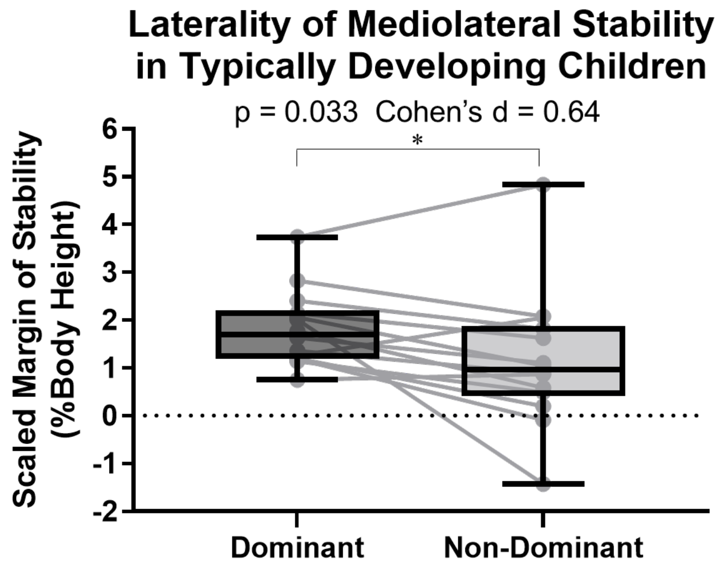

Figure 1.4: **Asymmetry between limbs for minimum lateral margin of stability for children with typical development**. Data related to previous publication [41]. Limb dominance determined by preferred kicking limb. Limb mean and 5 to 95 percentiles shown with box plots. Individual participant data shown with light gray circles. Participants were significantly less stable in the frontal plane when supporting weight with the non-dominant limb.

Laterality, defined here as the "existence of limb dominance" [85] or the presence of asymmetry rather than symmetry, has been detailed previously in the upper extremities showing between-arm differences in preference, specialization, or adaptation for some tasks [86–95]. Laterality may reflect limb-specific specialization, or "dynamic dominance", with the dominant limb prioritizing predictive control and the non-dominant limb prioritizing impedance control [88,92–96]. Predictive control involves developing accurate movement patterns to execute tasks. Impedance control incorporates feedback to adjust movement patterns to reduce errors and stabilize the position in the face of unpredictable or unexpected dynamic conditions. Laterality allows for more specific optimization and precision of movements than non-specialized movement organization, thus facilitating a greater range and depth of proficient movements [90,92,93]. The findings of these studies in upper-extremity function invite a better description of lower-extremity function than the common single-limb dominance approach.

During gait, one limb may prioritize stability while the other limb prioritizes mobility [85,97]. Under this hypothesized specialization of limbs, the stability limb would provide a greater contribution to body support while the mobility limb would contribute more to propulsion and movement trajectories. Other studies have investigated this functional asymmetry, finding evidence to support [84,85,97–100] and/or refute [97–103] this theory of lower-limb laterality. If such laterality exists, perhaps the stability and mobility priorities respectively coincide with the impedance and predictive control model hypothesized for the upper extremities. If limb laterality underlies the asymmetry with which stability is maintained during gait, then we would expect that degree of laterality to persist across tasks that challenge stability (impedance control) and mobility (predictive control). Four proposed “domains” of balance (Figure 1.5) represent a spectrum of quasi-static to dynamic tasks, including gait, that alter the priorities of stability and mobility. Task performance across domains may not be strongly correlated [104–107], likely due to the different demands of mobility and stability. However, we anticipated that the degree of laterality, which may reflect the stability/mobility priorities of the limb, would persist across these tasks of differing demands.

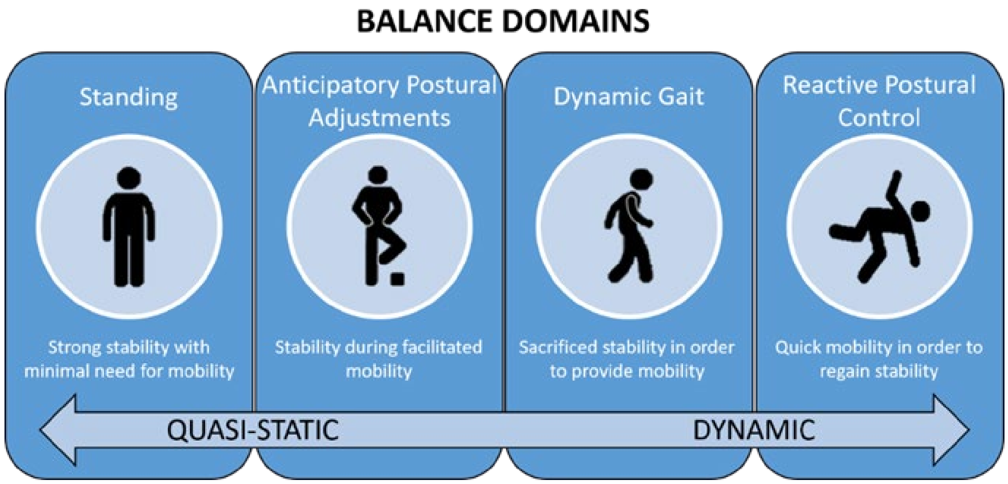

Figure 1.5: **Representation of the four primary balance domains.** Four primary balance domains relevant to postural control have been identified representing a spectrum of quasi-static and dynamic tasks: standing, anticipatory postural adjustments, dynamic gait, and reactive postural control [108–110]. Standing – maintaining control of the whole-body center of mass within the base of support while standing. This control is done by making small, continuous adjustments that control the center of mass to counteract the destabilizing force of gravity [110,111]. Anticipatory Postural Adjustments – postural control while self-initiating movements. In preparation for movement, predictive adjustments are made that will maintain stability and support mobility [106,110–112]. An example of this control is in the shifting of weight while standing towards one limb in preparation to take a step with the contralateral limb. Dynamic Gait – controlling stability and mobility while in motion. This complex coordination involves implementing proactive adjustments to impending perturbations (i.e. feedforward control) and modifying those responses to the environment (i.e. feedback control) [106,110,111]. Reactive Postural Control – coordinated whole-body responses to external perturbations. To prevent a fall, this requires rapid responses to perturbations of initially unknown magnitude or direction and can be completed with feet-in-place or compensatory stepping responses [110,111].

Stability and mobility are logical, functional targets for addressing the risk of falling and barriers to physical activity. As evident by the domains of balance (Figure

1.5), stability and mobility are interrelated and often competing in daily tasks. We do not know how these competing demands are negotiated. Laterality, also called sidedness or limb-dominance, is one construct that may provide insight on how these competing demands are addressed in motor tasks [113,114]. The results of this second aim, presented in Chapter 3, contributed to our understanding of functional lower-extremity laterality [84,85,114,97–103,113], including the theorized "stability" and "mobility" roles of each limb [84,85,97,98,102], and provided insight on limb function in roles where priorities are ambiguous (e.g. such as walking). Beyond the common definitions of "dominant" and "non-dominant" limbs, these functional definitions of limb laterality would provide a more informed approach to rehabilitation from injuries or impairments that exaggerate asymmetry. Stroke survivors, persons with cerebral palsy, lower-limb amputees, leg-length discrepancy, post-ACL reconstruction surgery, and persons with Parkinson's disease are a few examples of populations that have a higher risk of falling and/or present with asymmetrical function [41,115–126]. Although there are many populations that operate with inherent asymmetry in daily lower-extremity function, only a limited amount of literature, with mixed conclusions, is available regarding lower-extremity laterality in stability, especially across the four proposed balance domains [116,122,123,125,127]. Limb lateralization is typically characterized by questionnaires of self-initiated movement tasks [128,129]. These questionnaires, however, often do not span multiple domains of balance, including standing, anticipatory postural adjustments, dynamic gait, and reactive postural control [108–110].

The second aim, presented in Chapter 3, built on previous studies by investigating stability and the relationships to limb asymmetry, preference, and function spanning the four hypothesized balance domains [108–110]. This contribution was innovative because it was the first to assess limb laterality across balance domains, assessing the construct with limb-preference observations, performance evaluations, and objective, precise biomechanical measures. Additionally, these results will inform future study protocols and data interpretation for populations with and without impairments. Previous gait studies have collected data from only one limb [81–83] or pooled data between limbs [78–80]; however, these practices may exclude data informing functional conclusions. Asymmetry has commonly been a sign of impairment [85,130–132] but may offer an incomplete explanation of the observed motion. In addition to between-limb differences in function, asymmetry could be a result of limb-specific control priorities. These results provide additional perspective on how to interpret between-limb differences.

For our second aim, presented in Chapter 3, we used an innovative approach to understand gait stability and lower-extremity asymmetry. This aim was an important first step for building a framework for fall-risk and asymmetrical lower-extremity function and how gait stability is related to lower-extremity asymmetry in tasks across the balance domains.

1.3 Purpose and Hypothesis

This dissertation represents two basic studies that advanced our understanding of walking stability. The purpose of the first aim of this dissertation, presented in Chapter 2, was to investigate the possibility of proactive modifications to stability using the threat of large anterior and posterior perturbations and walking at multiple speeds. By addressing these novel aspects, we took important first steps towards determining the extent to which anterior and posterior walking stability are modifiable rehabilitative targets. Using anterior and posterior walking stability as modifiable rehabilitative targets will allow us to evaluate balance-training protocols for their impact on these stability measures and identify gait-retraining methods that modify fall risk without decreasing walking speed. This knowledge will also allow us to develop new protocols to train proactive adjustments to provide greater stability should a perturbation occur. We hypothesized that anteroposterior stability would be a modifiable aspect of gait. We predicted that unimpaired participants would increase stability protective against a loss of stability when threatened with large perturbations and would display more pronounced changes in stability at slower speeds. We expected that increases in stability would be accomplished by increasing step width, increasing step frequency, decreasing step length, and/or altering body orientation relative to the base of support.

The purpose of the second aim of this dissertation, presented in Chapter 3, was to establish the relationships between the laterality of gait stability and lower-extremity function across balance domains. This was an important first step in building a framework for fall-risk and asymmetrical lower-extremity function. The results from our investigation of limb laterality will contribute additional evidence to the discussion of functional lower-extremity laterality and the theorized “stability” and “mobility” roles of each limb [84,85,97,98,102]. These results will also provide insight on limb function in roles where priorities are ambiguous such as walking, when the limb roles are switched such as stepping with the non-preferred limb, and during other asymmetrical movement tasks such as turning. We hypothesized that asymmetry in walking would relate to asymmetry during tasks in other balance domains providing evidence of lower-extremity laterality. We predicted that unimpaired participants would exhibit significant correlations between walking stability asymmetry and other balance domain task asymmetries.

Chapter 2

PROACTIVE MODIFICATIONS TO WALKING STABILITY UNDER THE THREAT OF LARGE ANTEROPOSTERIOR PERTURBATIONS

2.1 Introduction

Falls while walking represent 33-71% of reported falls across many populations, such as stroke survivors, individuals with Parkinson's disease, and young, middle-aged, or older adults [17–24]. We propose that, from a biomechanical perspective, the ability of an individual to prevent a fall after a walking perturbation is influenced by two intrinsic factors: (1) the mechanical stability of unperturbed walking (i.e. the "initial conditions" before a perturbation) and (2) balance reaction capabilities after a perturbation (i.e. the "recovery skill" after a perturbation). A "stable" gait is one in which a relatively large perturbation is needed to initiate a loss of balance because the initial conditions prior to a perturbation are controlled and/or the person has the capacity to regain stability after a perturbation. Given a direct, logical relationship to fall risk, gait stability is a relevant target for interventions to reduce falls in at-risk populations. This chapter focuses on the first intrinsic factor of gait stability by investigating proactive modifications to walking stability.

We previously compared the anterior margin of stability (MoS) during unperturbed, overground walking of children with and without cerebral palsy. This previous study identified no differences in the anterior margin of stability (MoS) during walking (i.e. a proactive response) [41], despite those children with CP exhibiting an impaired response to anterior perturbations (i.e. a reactive response)

[26]. In other words, the children with CP did not walk in a manner that was proactively protective against a forward loss of balance. This result led us to question if, in the absence of neuromuscular impairment and with fully developed gait and balance, anteroposterior stability *can* be proactively modulated. For anteroposterior gait stability to be a target for fall-prevention intervention it must be modifiable.

The purpose of this study was to investigate whether participants, in the absence of neuromuscular impairment and with fully developed gait and balance, could proactively modify anteroposterior walking stability prior to perturbations when threatened with large anterior or posterior perturbations within a given walking speed. This investigation used large anterior and posterior perturbations, a focus on pre-perturbation steps, and multiple walking speeds to explore how anteroposterior stability can be modified proactively under threats to stability. A range of walking speeds will provide a greater framework with which to interpret proactive gait stability within multiple walking speeds. We hypothesized that anteroposterior stability would be a modifiable aspect of gait. We predicted that unimpaired participants would increase stability protective against a loss of stability when threatened with large perturbations and would display more pronounced changes in stability at slower speeds.

2.2 Methods

2.2.1 Participants

The University of Delaware IRB approved this study (Appendix D), and fourteen young adults, ages 20-36 years, provided informed consent to participate. Participants included a convenience sample of seven females and seven males (Table 2.1) who had no self-reported neurological or musculoskeletal disorders; no recent neural, muscular, or skeletal injuries; and no movement impairments. Additional participant descriptions included in Appendix A (Table A.1).

Table 2.1: **Description of participants.**

Descriptor	Sex	n	Mean	SD
Age (years)	Females	7	25.6	4.0
	Males	7	26.6	5.7
	All	14	26.1	4.8
Height (cm)	Females	7	170.8	4.4
	Males	7	184.1	8.2
	All	14	177.4	9.4
Mass (kg)	Females	7	63.7	9.1
	Males	7	72.4	11.5
	All	14	68.0	10.9
Body Mass Index ($kg \cdot m^{-2}$)	Females	7	21.8	2.8
	Males	7	21.2	1.9
	All	14	21.5	2.3

2.2.2 Protocol

Study participation included one visit to the KAAP Biomechanics Laboratory at the University of Delaware STAR Health Sciences Complex. After receiving informed consent and ensuring inclusion/exclusion criteria were met, we recorded the participant's sex, age, height, and mass. Foot dominance was determined using the revised version of the Waterloo Footedness Questionnaire [128] (Appendix A Figure A.1). The Waterloo Footedness Questionnaire involves 10 questions, 5 related to stability tasks and 5 related to mobility tasks, that the participant self-reports the extent to which they would use one limb or the other to perform the described activity on a 5-point scale. The cumulative score describes the participant's footedness (range: -2 = left always to 0 = no dominance to +2 = right always).

During all testing, the participant's movement was recorded with motion capture technology (Qualisys, Gothenburg, Sweden, 120 Hz) using a 41-point whole-body reflective marker set. All participants were equipped with a safety harness system attached to an overhead rail to arrest a fall, should any occur, prior to the knees or hands touching the treadmill or floor. This system also had an in-series strain gauge (Dillon, Fairmont, MN, USA, 100 Hz) to measure the support received by a participant. Harness support was classified as a failed recovery or fall if the participant measured a peak force surpassing 20% of their body weight [133]. If the participant fell or was pushed off the back of the treadmill, the session was paused until the participant returned to a standing position on the treadmill and was ready to continue.

Participants completed a five-minute walking warmup at an estimated preferred walking speed of 0.8 statures/s [134,135]. This period served as a physical warmup to prepare for the demands of the protocol tasks as well as a familiarization period with the testing equipment. The warmup and protocol tasks were completed while the participant walked on a computer-controlled treadmill (ActiveStep®, Simbex, Lebanon, NH, USA).

The nine protocol tasks included combinations of three scaled walking speeds (scaled to participant's height) and three perturbation types. Each participant walked at an estimated preferred walking speed of 0.8 statures per second ($stats \cdot s^{-1}$) [134,135], and at one slower and at one faster walking speed of 0.6 and 1.0 $stats \cdot s^{-1}$, respectively. Each speed was accompanied by one of three perturbation types: no perturbations, anterior perturbations (i.e. simulated trips), or posterior perturbations (i.e. simulated slips). A perturbation consisted of occasional (every 12 ± 2 steps) computer-controlled, rapid treadmill-belt accelerations during walking. The participant was informed of the treadmill belt speed and of the perturbation type before beginning the trial, but they were blind to the timing of the perturbations during the trial. Participants completed all nine combinations of walking speed and perturbation type in a different random order. After each trial, a subset of seven participants self-reported their perceived trial difficulty using a visual analog scale (Appendix A Figure A.4) [136]. After trials with perturbations, this same subset of participants self-reported their perceived change in difficulty over time using a Likert scale (Appendix A Figure A.5).

The treadmill is capable of delivering perturbations relative to initial foot contact during gait (Figure 2.1). Anterior perturbations refer to perturbations requiring forward recovery steps (i.e. simulated trips), and posterior perturbations requiring backward recovery steps (i.e. simulated slips). While the participant was walking on the treadmill at the pre-determined belt speed, perturbations were delivered every 10 to 14 steps (12 ± 2). The ActiveStep® software delivers anterior perturbations approximately at mid-swing by delaying the delivery by 0.2 s after foot strike detection. These perturbations were similar to those that we've previously applied to young adults [30] with a peak perturbation velocity of 1.8 $m \cdot s^{-1}$ achieved in 0.12 s (displacement = 0.22 m, acceleration = 15.00 $m \cdot s^{-2}$). After the perturbation, participants needed one or more steps to recover while returning to walking at the treadmill belt speed for that trial. Posterior perturbations occurred shortly after initial foot contact. These perturbations were also similar to those that we've previously applied to young adults [29] with a peak perturbation velocity of -0.30 $m \cdot s^{-1}$ achieved in 0.14 s (displacement and acceleration dependent on initial belt velocity). Following the perturbation, the treadmill resumed the predetermined belt speed requiring the participant to resume walking at the target speed prior to the next perturbation. Each combination of speed and perturbation type consisted of three minutes of walking. We aimed to use large, challenging perturbations to encourage proactive modifications to stability. Small perturbations may not require proactive modifications to stability; too large of perturbations may demotivate the participant and hinder the feasibility of the protocol due to an inability to recover without falling and lengthened protocol durations from falls. The perturbation sizes were intended to be challenging to these participants in order to create a credible threat to stability, but not to be large enough to consistently elicit a fall into the safety harness. Two-minute rest periods occurred between each trial, and were extended to allow each participant to recover as needed.

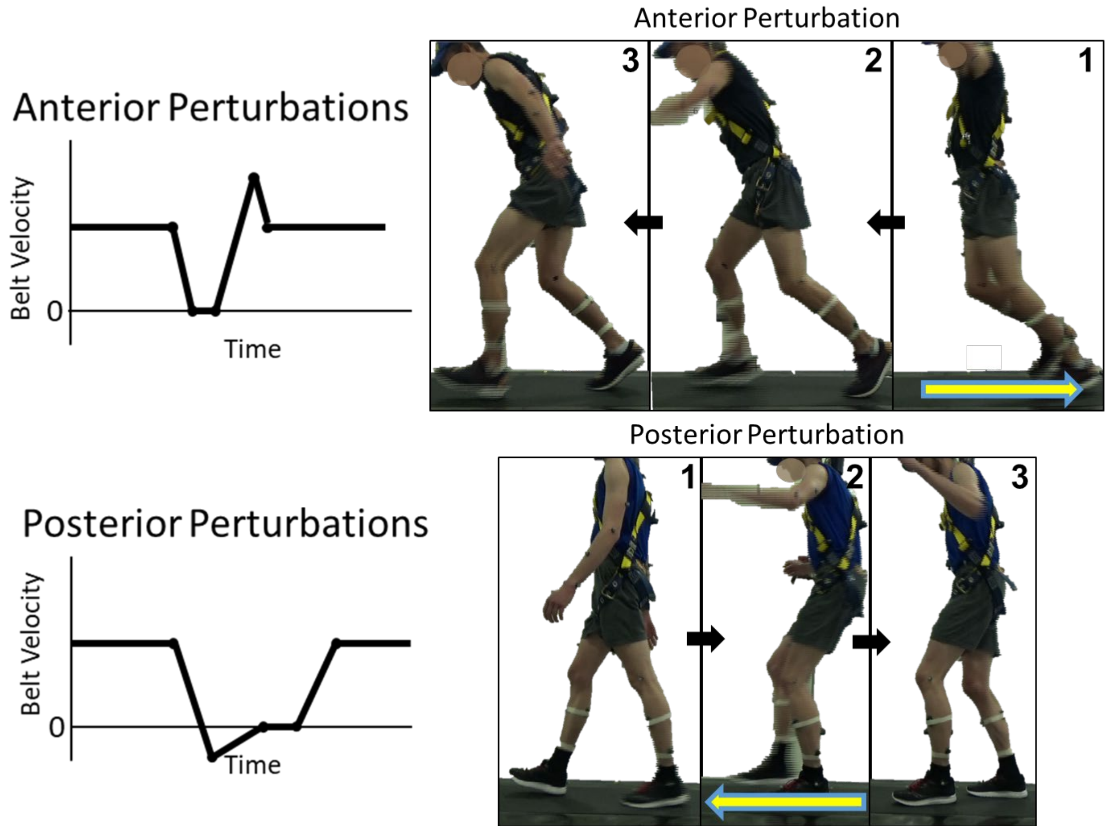

Figure 2.1: **Example profiles of anterior and posterior perturbations.** Anterior perturbations represent simulated trips, and posterior perturbations represent simulates slips. Positive and negative belt velocities correspond to treadmill belt progressions resulting in forward or backward stepping, respectively. Top Row – a time series and images of an anterior perturbation, progressing from right to left, that occurred approximately at mid-swing. Bottom Row – a time series and images of a posterior perturbation, progressing from left to right, that occurred at foot strike.

2.2.3 Analysis

The final right and left step prior to each perturbation were evaluated for each trial. For the three no perturbation conditions, sequential right and left steps were evaluated every 10-15 seconds of each trial. We used Visual 3D (C-Motion, Inc., Germantown, MD, USA, version 2021) to filter the motion capture data (4^{th} order lowpass Butterworth filter with a 6 Hz cutoff) and determine the whole-body center of mass. Foot-strike and toe-off events were determined using the "coordinate-based treadmill algorithm" described by Zeni and colleagues [137]. Mid-swing was determined as the first frame where the swing limb toe marker passed anterior to the stance limb toe marker, a point where a trip or stumble is likely to occur [76]. Anteroposterior stability was quantified using the MoS [35] at mid-swing and at foot strike for each analyzed step using custom Excel Visual Basic for Applications software (Microsoft, Redmond, WA, USA, version 2016). The MoS was adapted to account for the velocity of the treadmill belt [30] (Figure 2.2) and scaled to the participant's height [41,138]. At foot strike, when a simulated slip or posterior perturbation could occur, we measured the posterior margin of stability. The MoS was calculated as the distance between the extrapolated center of mass (see Figure 2.2) and the stepping limb heel representing the posterior edge of the stepping limb base of support (Equation 2.1):

$$MoS_{FS} = xCoM - BoS, \qquad \text{Equation 2.1}$$

where MoS_{FS} represents the MoS at foot strike, xCoM represents the anteroposterior position of the extrapolated center of mass, and BoS represents the anteroposterior position of the posterior edge of the base of support. Given the global coordinate system of the laboratory, Equation 2.1 modified the order of the terms on the right side of the equation from Hof and colleagues proposed equation [35] so that a positive MoS_{FS} value indicated a stable position relative to a slip (i.e. a posterior loss of balance). A positive value indicated that the extrapolated center of mass was located anterior to the posterior edge of the stepping limb's base of support. At mid-swing, when a simulated trip or anterior perturbation could occur, we measured the anterior margin of stability as illustrated in Figure 2.2. The MoS was calculated as the distance between the extrapolated center of mass and the stance limb toe representing the anterior edge of the stance limb base of support (Equation 2.2):

$$MoS_{MS} = BoS - xCoM, \qquad \text{Equation 2.2}$$

where MoS_{MS} represents the MoS at mid-swing, BoS represents the position of the anterior edge of the base of support, and xCoM represents the position of the extrapolated center of mass. A negative value indicated that the extrapolated center of mass was located anterior to the anterior edge of the stance limb's base of support, an unstable position relative to a trip (i.e. an anterior loss of balance). Mean values were calculated for each limb within a condition as the primary outcome measure. A repeated-measures factorial ANOVA was conducted in SPSS (IBM, Armonk, NY, USA, version 28) to evaluate the main effects and interactions of the reference limb (dominant or non-dominant), walking speed (slow, estimated preferred, or fast), and perturbation type (simulated trips, none, or simulated slips). Significance was set at $p < 0.05$, and effect sizes were reported using partial eta squared values (small effect size: $\eta^2 < 0.06$; medium: $0.06 < \eta^2 < 0.14$; large $\eta^2 > 0.14$). Conservatively assuming independence between conditions, 12 participants would have provided 80 percent power to detect a medium effect ($\eta^2 = 0.05$) as significant.

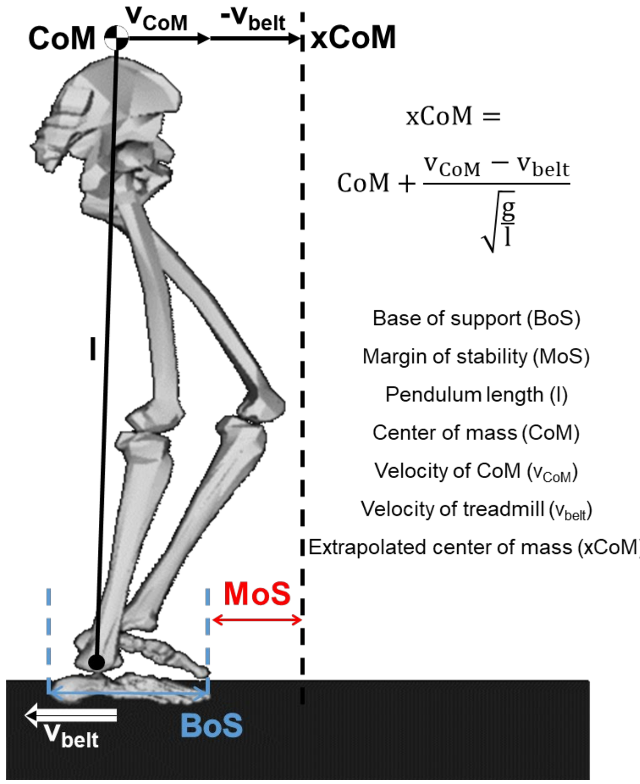

Figure 2.2: **Illustration of the anteroposterior margin of stability at mid-swing accounting for treadmill belt velocity.** Figure modified from previous publication [41]. The margin of stability represents the distance between the base of support and the extrapolated center of mass (center of mass position + scaled center of mass velocity + treadmill belt velocity). Positive values represent a state of stability (i.e. the extrapolated center of mass is within the base of support or advantageously placed away from the edge of the base of support), and a perturbation is needed to initiate a fall in that direction. Negative values represent a state of instability (i.e. the extrapolated center of mass is outside the base of support), and a compensatory action such as taking a step, applying an external force, or counter-rotating segments about the center of mass is needed to prevent a fall.

2.3 Results

2.3.1 Participants and Perturbations

Three participants, one female and two males, were excluded from the analysis due to incomplete protocols (partial file corruption, mechanical error with the treadmill, and elective end to participation due to a heightened level of excitement/nervousness related to the unknown timing of perturbation onset). Exclusion due to heightened levels of excitement/nervousness was an anticipated risk, and our rate was similar to that of a previous project [25]. Of these remaining 11 participants, all reported right limb dominance according to their preferred kicking limb. The average Waterloo Footedness score (possible range -2 to +2) was 0.65 with a standard deviation of 0.54 and range of -0.30 to 1.60 (Figure 2.3, Appendix A Figure A.1). The number of steps analyzed per combination of walking speed and perturbation condition is shown in Figure 2.4. With each left and right step counting individually, an average of 253.3 total steps (SD: 11.0; Range: 238 to 276 steps) were analyzed per participant with an average of 177.1 pre-perturbation steps (SD: 8.2; Range: 166 to 190 steps). Participants received an average of 88.5 total perturbations (SD: 4.1; Range: 83 to 95 perturbations). Perturbation trials consisted of an average of 14.8 perturbations (SD: 0.7; Range: 12 to 20 perturbations).

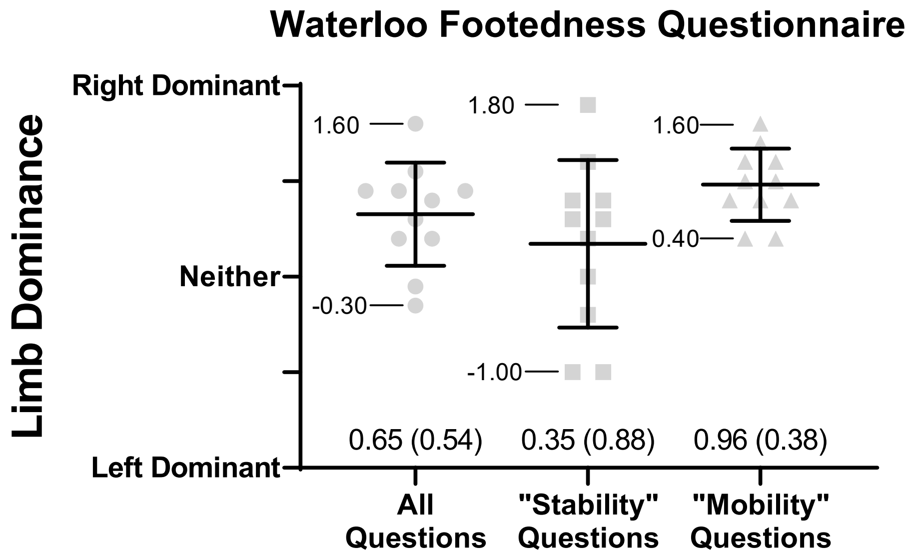

Figure 2.3: **Results from the Waterloo Footedness Questionnaire – Revised** [128]**.** Participants responded to ten questions, five related to "stability" tasks and five related to "mobility" tasks, with which limb they would perform a task (left, right, or equal) and the extent to which they would use that limb for the task (always or usually). Mean (standard deviation) and maximum and minimum values identified. See also Appendix A Figure A.1.

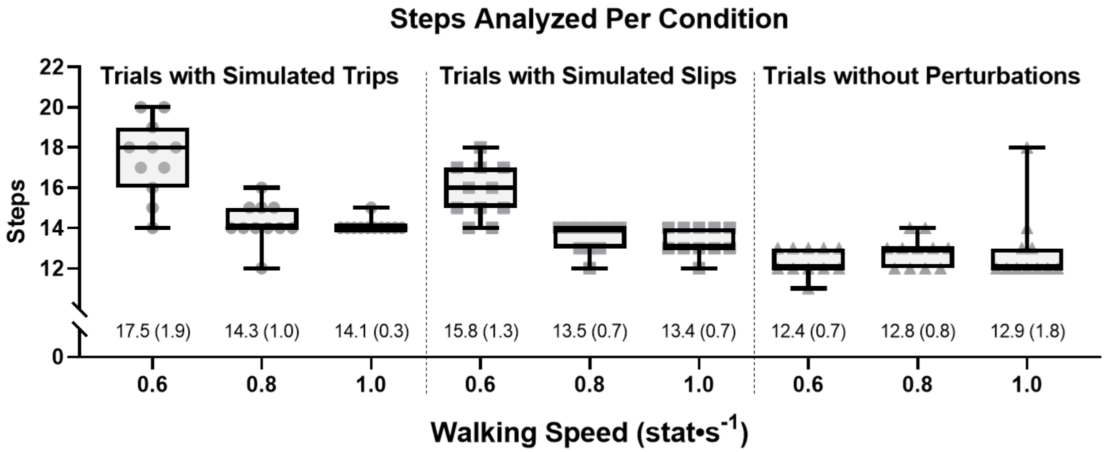

Figure 2.4: **Analyzed steps per combination of walking speed and perturbation condition**. Summary data shown with median and 5 to 95 percentile box plots and mean (standard deviation) values. Individual participant data shown with light gray circles (simulated trips), squares (simulated slips), and triangles (no perturbations). The final right and left steps prior to each perturbation were analyzed. For trials without perturbations, sequential right and left steps were sampled every 10-15 seconds. The number of steps represents the number of left and right steps included in the analysis (i.e. 16 represents 16 left steps and 16 right steps). The number of steps also represents the number of perturbations experienced within that condition (i.e. 16 steps represents 16 perturbations).

2.3.2 Posterior Margin of Stability at Foot Strike

The posterior MoS at foot strike represents stability relative to a posterior loss of balance, such as that during a slip. These reported results prioritize the comparison of the no perturbation condition to the posterior perturbation condition. Neither the three-way interaction of perturbation type, walking speed, and stepping limb was significant ($p = 0.693$; $\eta^2 = 0.042$), nor were there significant two-way interactions ($p > 0.062$, $0.075 < \eta^2 < 0.197$). While the main effect of stepping limb was not significant ($p = 0.895$, $\eta^2 = 0.002$), the main effects of perturbation type ($p < 0.001$, $\eta^2 = 0.766$) and walking speed ($p < 0.001$, $\eta^2 = 0.991$) were significant (Figure 2.5). Post-hoc comparisons of the perturbation trials (simulated trips and simulated slips) to trials with no perturbations showed that the posterior MoS at foot strike was significantly more positive during trials with posterior perturbations ($p < 0.001$, mean difference (standard error) 1.70 (0.26) %BH, Figure 2.6A) and during trials with anterior perturbations ($p < 0.001$, 1.13 (0.21) %BH, Figure 2.6A).

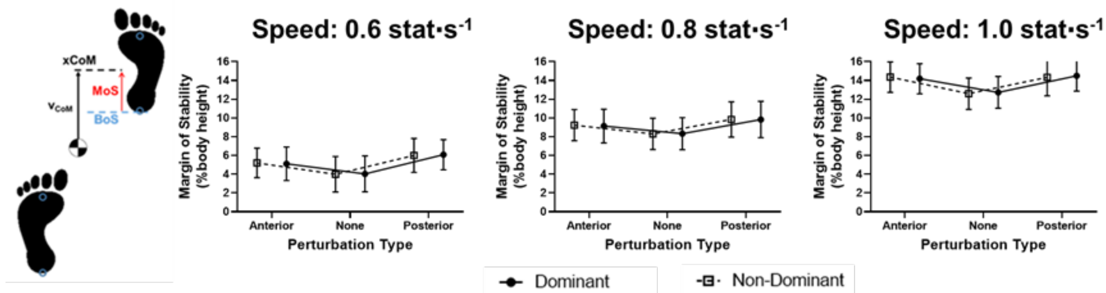

Figure 2.5: **Summary results for the posterior margin of stability at foot strike.** Participants proactively modified their stability with the threat of perturbations to walk in a way that increased their stability relative to a posterior loss of balance. With each increase in walking speed, participants increased their stability relative to a posterior loss of balance. Compared to trials with no perturbations, the posterior MoS at foot strike was significantly more positive during trials with posterior perturbations ($p < 0.001$, mean difference (standard error) 1.70 (0.26) %BH) and during trials with anterior perturbations ($p < 0.001$, 1.13 (0.21) %BH). The dominant limb is shown with solid circles and solid lines. The non-dominant limb is shown with open squares and dashed lines.

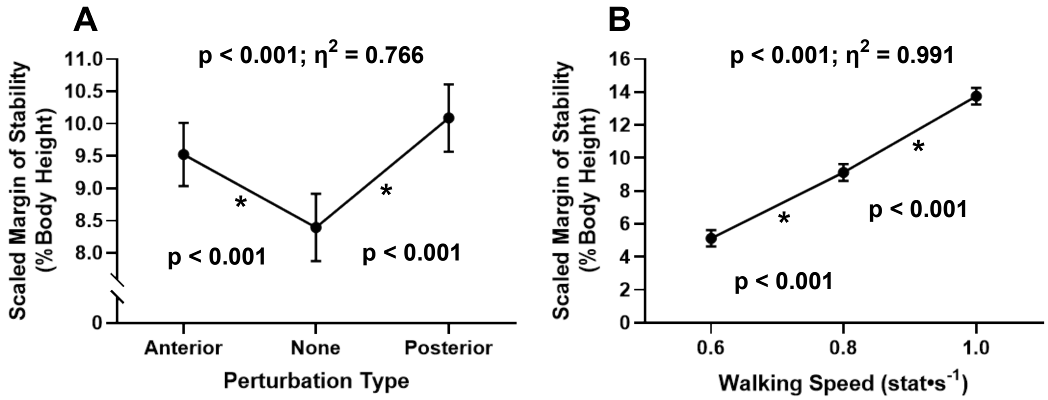

Figure 2.6: **Posterior margin of stability at foot strike.** Estimated marginal means and standard errors of posterior margin of stability across perturbation types (simulated trips, none, or simulated slips) and walking speeds (slow, estimated preferred, or fast). <u>Panel A</u> – Participants showed increased posterior stability (i.e. more positive MoS) at foot strike relative to a posterior loss of balance during trials with perturbations compared to trials without perturbations. <u>Panel B</u> – With each increase in walking speed, participants increased their stability relative to a posterior loss of balance.

2.3.3 Anterior Margin of Stability at Mid-Swing

The anterior MoS at mid-swing represents stability relative to a forwards loss of balance such as that during a trip. These reported results prioritize the comparison of the no perturbation condition to the anterior perturbation condition. The three-way interaction of perturbation type, walking speed, and stepping limb was not significant ($p = 0.067$; $\eta^2 = 0.193$). The two-way interaction of walking speed and stance limb was also not significant ($p = 0.371$; $\eta^2 = 0.094$), but the two-way interactions of perturbation type and stance limb ($p = 0.006$; $\eta^2 = 0.398$) and perturbation type and walking speed ($p = 0.008$; $\eta^2 = 0.285$) were significant (Figure 2.7). Post-hoc comparisons of the perturbation types (simulated trips, none, or simulated slips) and stance limb (dominant or non-dominant) interactions showed no significant differences between perturbation types when stance was with the non-dominant limb ($p > 0.995$, mean difference (standard error) < 0.06 (0.26) %BH), but significant differences between all perturbation types when stance was with the dominant limb ($p < 0.032$). When stance was with the dominant limb, the anterior MoS at mid-swing was significantly more negative (i.e. more unstable relative to an anterior loss of balance such as a trip) during trials with anterior perturbations ($p = 0.005$, mean difference (standard error) 0.63 (0.15) %BH), and also for trials with posterior perturbations ($p = 0.001$, 0.90 (0.17) %BH), compared to trials with no perturbations (Figure 2.8A).

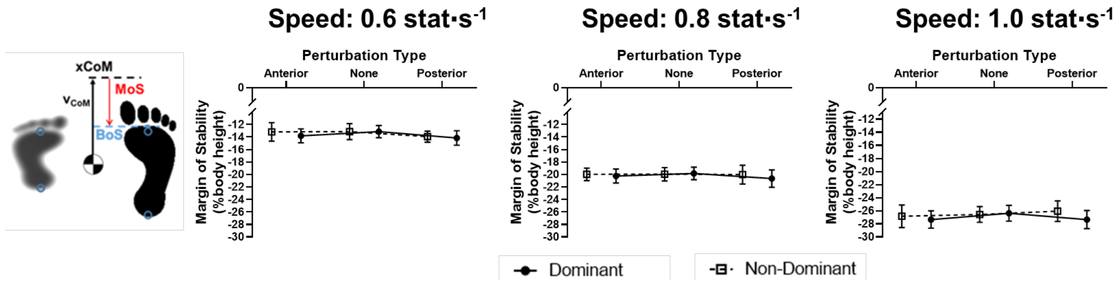

Figure 2.7: **Summary results for the anterior margin of stability at mid-swing.** Participants showed a between-limb asymmetry in proactive stability modifications where participants were less stable anteriorly during stance on the dominant limb compared to trials without threats to stability and compared to stance on the non-dominant limb. When stance was with the dominant limb, the anterior MoS at mid-swing was significantly more negative (i.e. more unstable relative to an anterior loss of balance such as a trip) during trials with anterior perturbations ($p = 0.005$, mean difference (standard error) 0.63 (0.15) %BH), and also for trials with posterior perturbations ($p = 0.001$, 0.90 (0.17) %BH), compared to trials with no perturbations. There was no significant difference between stance limbs for the no perturbation type ($p = 0.309$), but significant differences between stance limbs during trials with anterior perturbations ($p = 0.027$, 0.48 (0.18) %BH), and also for trials with posterior perturbations ($p = 0.016$, 0.71 (0.25) %BH), where the anterior MoS at mid-swing was significantly more negative for stance on the dominant limb. The dominant limb is shown with solid circles and solid lines. The non-dominant limb is shown with open squares and dashed lines.

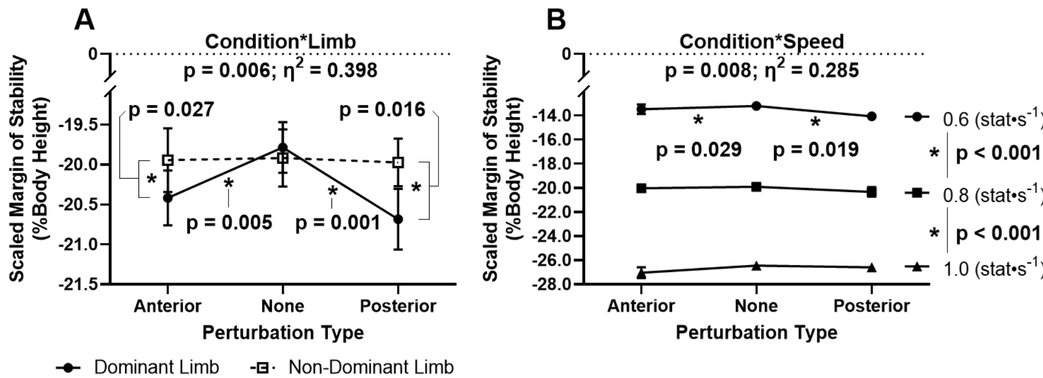

Figure 2.8: **Anterior margin of stability at mid-swing.** Estimated marginal means and standard errors of anterior margin of stability interactions of perturbation types (simulated trips, none, or simulated slips) with stance limb (dominant or non-dominant) and perturbation types with walking speeds (slow, estimated preferred, or fast). <u>Panel A</u> – Participant's had less anterior stability (i.e. more negative MoS) during stance on the dominant limb for trials with perturbations compared to trials without perturbations. Decreased anterior stability during stance on the dominant limb compared to stance on the non-dominant limb was present within perturbation trials, but not within the no perturbation trials. The dominant limb is shown with solid circles and solid lines. The non-dominant limb is shown with open squares and dashed lines. <u>Panel B</u> – A decrease in anterior stability was observed between trials with and without perturbations within the slow walking speed condition. With each increase in walking speed, participants increased their instability relative to a forward loss of balance.

A between-limb asymmetry in stability was also observed within trials with perturbations that was not present for trials without perturbations. Post-hoc comparisons between stance limbs within each perturbation type showed no significant difference between stance limbs for the no perturbation type ($p = 0.309$), but significant differences between stance limbs during trials with anterior perturbations ($p = 0.027$, 0.48 (0.18) %BH), and also for trials with posterior perturbations ($p = 0.016$, 0.71 (0.25) %BH), where the anterior MoS at mid-swing was significantly more negative for stance on the dominant limb (Figure 2.8A).

A significant difference in stability was observed at the slow walking speed between trials with and without perturbations. Post-hoc comparisons between perturbation types within walking speeds showed no significant differences between perturbation types for the fast and estimated preferred walking speeds ($p > 0.053$, mean difference (standard error) < 0.59 (0.21) %BH, Figure 2.8B). There was a significant difference at the slow walking speed between the condition with no perturbations and with anterior perturbations ($p = 0.029$, 0.58 (0.18) %BH), and also with posterior perturbations ($p = 0.019$, 0.86 (0.25) %BH), where the anterior MoS at mid-swing was significantly more negative during trials with perturbations (Figure 2.8B).

2.3.4 Gait Parameters

Accompanying these proactive modifications to stability, in response to the threat of perturbations, were changes in step length and step rate, but not step width. For step length, there were significant main effects of perturbation type ($p < 0.001$, $\eta^2 = 0.676$) and walking speed ($p < 0.001$, $\eta^2 = 0.992$). Post-hoc comparisons of the perturbation types showed that, compared to trials with no perturbations, step length was significantly shorter during trials with anterior perturbations ($p = 0.005$, mean difference (standard error) 0.017 (0.004) m) and trials with posterior perturbations ($p = 0.001$, 0.022 (0.004) m, Figure 2.9A). Post-hoc comparisons of the walking speeds showed that walking faster increased step length (slow to estimated preferred: $p < 0.001$, 0.112 (0.004) m; estimated preferred to fast: $p < 0.001$, 0.102 (0.004) m, Figure 2.9A).

For step rate, there were significant main effects of perturbation type ($p < 0.001$, $\eta^2 = 0.672$) and walking speed ($p < 0.001$, $\eta^2 = 0.992$). Post-hoc comparisons of the perturbation types showed that, compared to trials with no perturbations, step rate was significantly higher during trials with anterior perturbations ($p = 0.004$, mean difference (standard error) 0.045 (0.010) $steps \cdot s^{-1}$) and trials with posterior perturbations ($p = 0.002$, 0.060 (0.012) $steps \cdot s^{-1}$, Figure 2.9B). Post-hoc comparisons of the walking speeds showed that walking faster increased step rate (slow to estimated preferred: $p < 0.001$, 0.216 (0.008) $steps \cdot s^{-1}$; estimated preferred to fast: $p <0.001$, 0.186 (0.008) $steps \cdot s^{-1}$, Figure 2.9B).

For step width, there were no significant main effects of perturbation type ($p = 0.105$, $\eta^2 = 0.202$, Figure 2.9C) or walking speed ($p = 0.319$, $\eta^2 = 0.108$, Figure 2.9C).

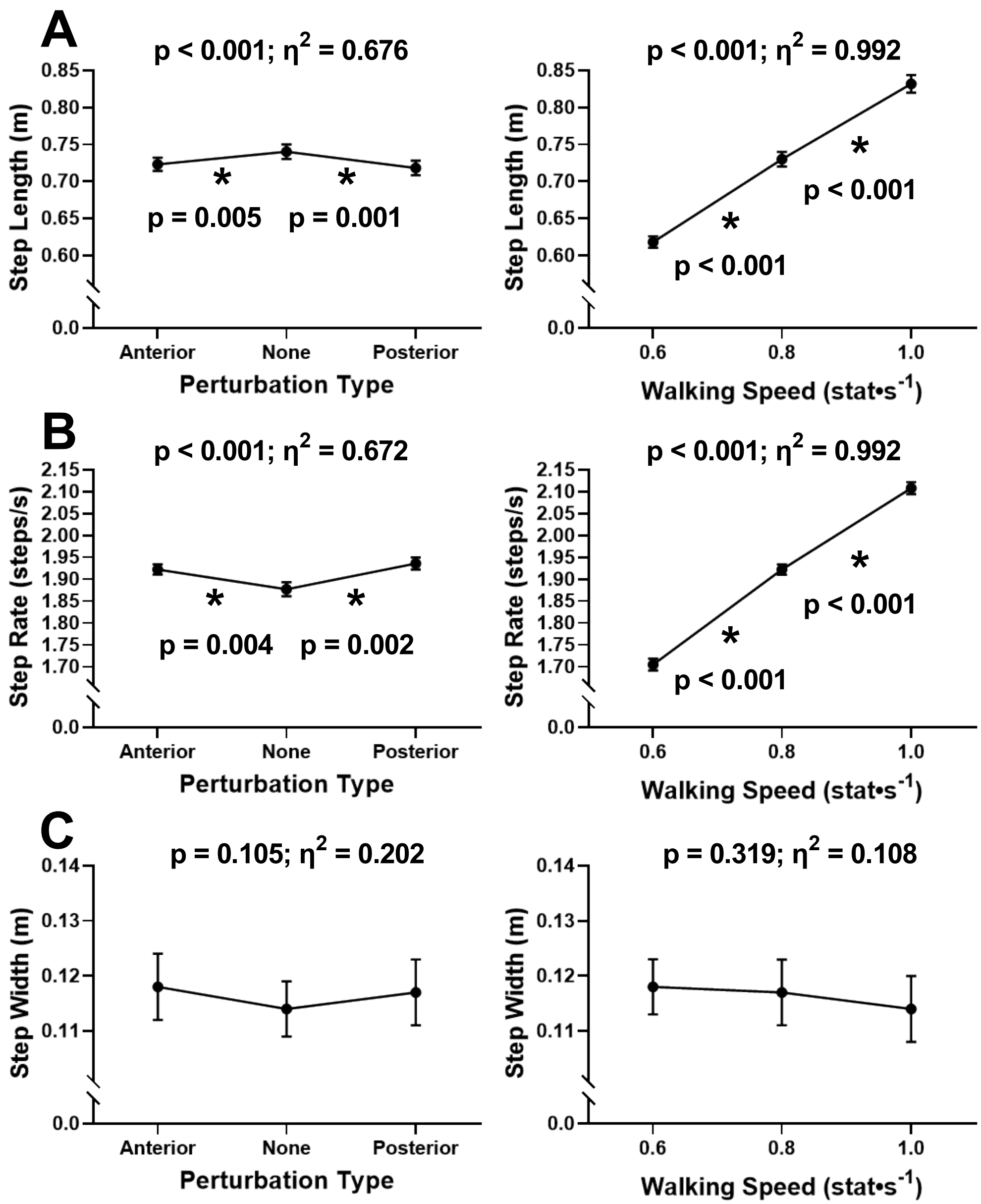
A
p < 0.001; η2 = 0.676
Step Length (m)
p = 0.005
p = 0.001
Anterior
None
Posterior
Perturbation Type
p < 0.001; η2 = 0.992
p < 0.001
p < 0.001
Walking Speed (stat•s-1)
B
p < 0.001; η2 = 0.672
Step Rate (steps/s)
p = 0.004
p = 0.002
Perturbation Type
p < 0.001; η2 = 0.992
p < 0.001
p < 0.001
Walking Speed (stat•s-1)
C
p = 0.105; η2 = 0.202
Step Width (m)
Perturbation Type
p = 0.319; η2 = 0.108
Walking Speed (stat•s-1)

Figure 2.9: **Gait parameters across perturbation types and walking speeds.** Estimated marginal means and standard errors of gait parameters across perturbation types (simulated trips, none, or simulated slips) and walking speeds (slow, estimated preferred, or fast). Row A – Participants decreased step length for trials with perturbations compared to trials without perturbations and increased step length with increasing walking speed. Row B – Participants increased step rate for trials with perturbations compared to trials without perturbations and increased step rate with increasing walking speed. Row C – Participants did not change step width when threatened with perturbations or when changing walking speeds.

There was also an accompanying modification in time spent in double support. There was a significant two-way interaction of condition and speed ($p = 0.004$, $\eta^2 = 0.311$). Post-hoc comparisons between perturbation types within walking speeds showed a significant difference within the slow walking speed between the condition with no perturbations and with posterior perturbations ($p = 0.042$, mean difference (standard error) 0.994 (0.335) %gait cycle) where the percent time spent in double support was significantly shorter for trials with the threat of posterior perturbations compared to no perturbations (Figure 2.10).

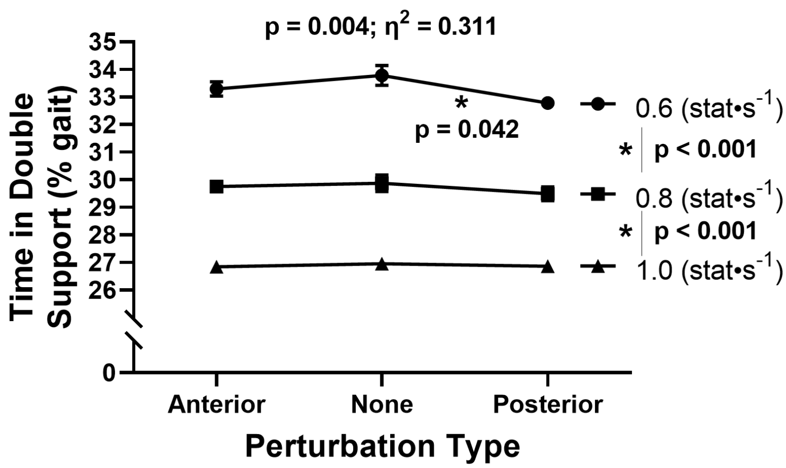

Figure 2.10: **Time spent in double support as a percentage of gait cycle across perturbation types and walking speeds.** Estimated marginal means and standard errors of gait parameters across perturbation types (simulated trips, none, or simulated slips). A decrease in the percent of time spent in double support was observed when threatened with posterior perturbations at the slow walking speed. With each increase in walking speed, participants decreased their percentage of time spent in double support.

MoS and gait parameter means and standard deviations for all combinations of perturbation type (simulated trips, none, or simulated slips), walking speeds (slow, estimated preferred, or fast), and stepping limb (posterior MoS at foot strike) / stance limb (anterior MoS at mid-swing) are reported in Appendix A (MoS: Table A.2; Gait parameters: Table A.5). All results from the factorial ANOVAs are presented in Appendix A (MoS at foot strike: Table A.3; MoS at mid-swing: Table A.4; Gait parameters: Table A.6). Summary figures showing step length, step rate, step width, stance time, swing time, and time spent in double support for all combinations of perturbation type, walking speeds, and stepping limb are reported in Appendix A Figure A.2 and Figure A.3.

2.4 Discussion

The purpose of this study was to investigate whether participants, in the absence of neuromuscular impairment and with fully developed gait and balance, could proactively modify stability prior to perturbations when threatened with large anterior or posterior perturbations within a given walking speed. At foot strike, when a slip could occur, participants significantly increased their stability relative to a posterior loss of balance when threatened with simulated slips (Figure 2.6A). At mid-swing, when a trip could occur, participants did not increase their stability relative to an anterior loss of balance when threatened with simulates trips. Participants were less stable relative to an anterior loss of balance when stance was on the dominant limb, but not the non-dominant limb, when threatened with simulated trips (Figure 2.8A). A previous study with some similarities to our protocol also observed modifications in margin of stability where healthy adult participants increased their stability relative to a posterior loss of balance with threats to stability [61]. However, this study used lateral perturbations, averaged values across pre- and post-perturbation steps, and walked at a single speed. This approach minimally threatens anteroposterior balance, does not isolate proactive modifications, and does not account for various walking speeds, thus limiting the comparisons to our data.

In response to the threat of perturbations, proactive modifications to stability were accompanied by a decrease in step lengths and an increase in step rates, but no change in step widths (Figure 2.9 A-C). Previous results for changes in gait in anticipation of overground (walking at a self-selected speed) or treadmill (walking at a fixed speed) slipping also demonstrated decreased step lengths [139,140] and increased step rates [140] in line with the adoption of a more 'cautious gait'. These results are also in agreement with previously reported outcomes showing decreased step lengths [43,60,61] and increased step rates [53,60,61] when responding to mediolateral perturbations while walking on a treadmill, although walking speed was also shown to decrease in one protocol using a self-paced treadmill [43], and walking slower changes step lengths and step rates. Other studies have also reported a responsive increase in step width [42,43,53,61] that we did not see. It is possible that the lack of a direct threat to lateral stability did not require the same kinematic adaptation to protect against a lateral loss of balance. One of these studies also incorporated anteroposterior perturbations and observed no difference in step length, frequency, or width [60], but the magnitude or intensity of the perturbations that were used is unclear, thus limiting the inferences to this work.

Interventions targeted at reducing fall risk may be able to improve walking stability within a given speed by promoting shorter, more frequent steps. Our results show that these adjustments are able to improve posterior stability at foot strike. This speed-specific adaptation is important as gait interventions often target beneficial increases in walking speed, a change that would correspond with less anterior stability. However, decreasing stride length to improve stability may be a challenge for participant groups who already present with shorter strides, such as persons with Parkinson's disease [141]. A meaningful argument against modifying stride lengths away from self-selected parameters is the likelihood that this change will decrease walking economy and increase the mechanical work done at the joints [142,143]. Walking stability, economy, and joint work are all important considerations for gait interventions in order to provide the best care for the individual.

In response to the threat of posterior perturbations at the slow walking speed, proactive modifications to stability were accompanied by a decrease in the percent of time spent in double support (Figure 2.10). This decrease in time was unexpected as double-support allows for effective center of pressure modulation protective against anteroposterior perturbations [144]. Returning to our proposed two intrinsic factors for fall prevention during walking (Figure 1.1), this result may be an example of our unimpaired participants relying on their capacity to regain stability after a perturbation (factor two) rather than improving their initial stability conditions (factor one). By decreasing the percent of time spent in double support, resulting in an increase in percent of time spent in single support, the participant would be in a position to respond more quickly to a perturbation by placing the swing limb back down after a posterior perturbation rather than needing to reposition a previously placed step. This strategy could be particularly useful for posterior perturbations as the direction of the recovery step is opposite the direction of walking. When walking at faster speeds, the need to maintain walking speed may not allow for this decrease in percent time spent in double support.

Interestingly, all of the proactive modifications to dynamic stability occurred to increase stability relative to a *posterior* loss of balance, even with the threat of anterior perturbations (Figure 2.6A and Figure 2.8A). Compared to anterior single- and multiple-stepping thresholds, posterior single- and multiple-stepping thresholds have resulted from smaller initial accelerations from treadmill-based perturbations in unimpaired adults [29]. Posterior stepping thresholds in older adults have been shown to differentiate between fallers and non-fallers [145] and in older adult women to be prospectively related to falls [25]. Increasing posterior but not anterior stability may be another example of our unimpaired participants relying on their capacity to regain stability after a perturbation (factor two) rather than modifying their initial stability (factor one). Posterior perturbations are in the opposite direction of walking and towards an environment outside of the forward or peripheral vision, which may increase the perceived threat of posterior perturbations. In contrast, anterior perturbations are collinear with the walking direction and towards a well-seen recovery space. Therefore, even with the threat of anterior perturbations, participants may have also protected against the greater perceived threat of a posterior loss of balance. Alternatively, gait modifications to maintain or improve anterior stability with the threat of anterior perturbations may have also benefited posterior stability, or may have resulted in improved anterior stability at points of the gait cycle other than at mid-swing.

A greater perceived threat of posterior perturbations compared to anterior perturbations is supported by self-reported data from a subset of our participants. After each trial, seven of our participants responded on a visual analog scale (Appendix A Figure A.4) [136] to the prompt "Make a mark indicating how difficult you perceived the trial you just completed" where the responses could range from Very Easy (scored 0) to Very Difficult (scored 100) (Figure 2.11). Within each walking speed, trials with posterior perturbations were, on average, perceived as more difficult than trials with anterior perturbations, and both perturbation types were perceived as more difficult than trials with no perturbations (Figure 2.11).

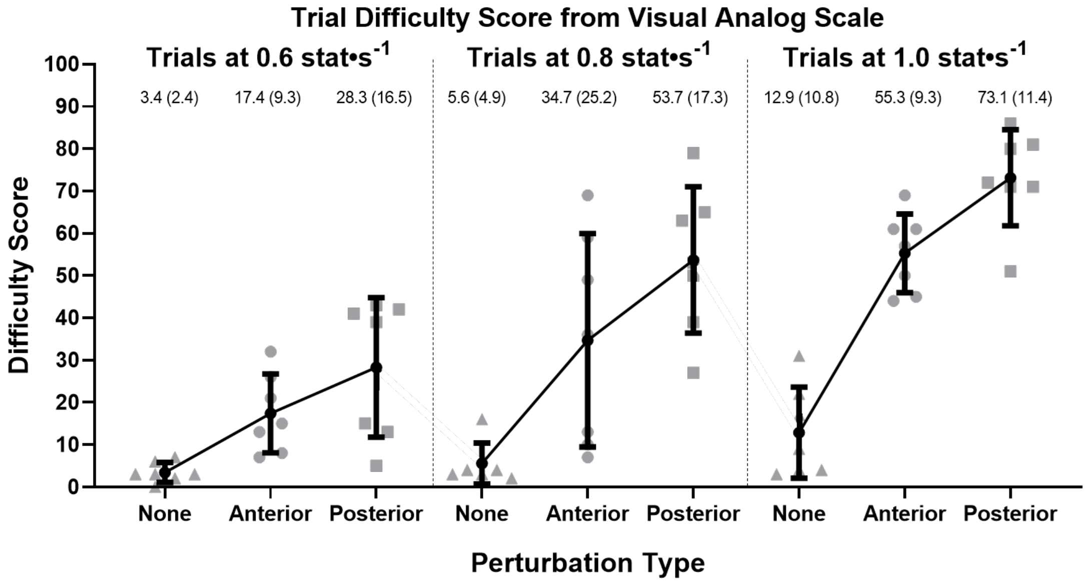

Figure 2.11: **Perceived trial difficulty.** Participant response mean and standard deviation represented with solid black circles and error bars. Individual responses shown with gray triangles (no perturbation trials), gray circles (anterior perturbation trials), and gray squares (posterior perturbation trials). Across all walking speeds, we observed trends suggesting that trials with posterior perturbations were perceived as more difficult than trials with anterior perturbations and all trials with perturbations were more difficult than trials without perturbations.

Another interesting finding was the between-limb asymmetry in MoS observed at mid-swing (Figure 2.8A) but not at foot strike. Trips are likely to pose a high risk of falling should the disturbance occur at mid-swing in the gait cycle [76,146]. At mid-swing, participants were less stable anteriorly during stance on the dominant limb compared to trials without threats to stability and compared to stance on the non-dominant limb. This result provides evidence for the proposed functional divide between the dominant limb prioritizing mobility and the non-dominant limb prioritizing stability [85,97], explored further in Chapter 3. Within this protocol, participants could not select with which limb to step as perturbations occurred during walking and could be triggered during stance on either limb.

Not addressed in this analysis was the potential for changes in recovery responses over time. These large anterior or posterior perturbations presented a novel walking challenge to participants. Because of this novelty, participants may change their proactive modifications over time as they develop a better understanding of the physical requirements needed to recover from the perturbation. After trials with perturbations, a subset of seven participants responded on a five-point Likert scale (Appendix A Figure A.5) to the prompt "During this trial, it became ______ to recover from the perturbations over time" where the responses could range from Much Easier (scored -2) to Much Harder (scored +2). Participants responded that the anterior and posterior perturbations were easier to recover from over time (Figure 2.12). This trend towards easier recovery over time suggests that there may be a learning effect occurring across the three-minute trial. Because participants did not report an increase in the difficulty over time, fatigue may not be a confounding factor for this protocol with unimpaired young adults. The pre-perturbation steps prior to the first perturbation or across the first minute of perturbations (representing approximately five perturbations) could explore the early responses that may better represent unsure gait in out-of-the-lab experiences and exclude the potential learning effect.

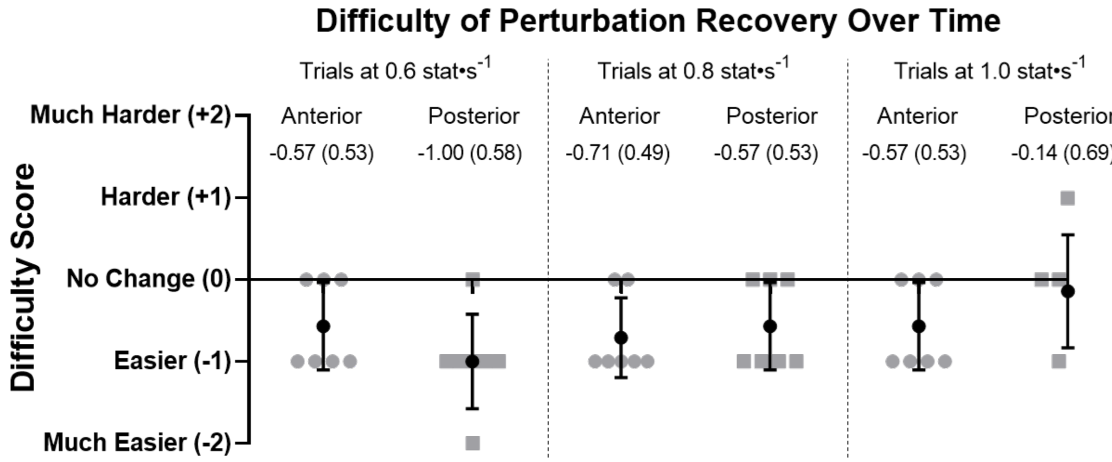

Figure 2.12: **Perceived change in difficulty of perturbation recovery over time.** Participant response mean and standard deviation represented with solid black circles and error bars. Individual responses shown with gray circles (anterior perturbation trials) and gray squares (posterior perturbation trials). Across all combinations of trials with perturbations and walking speeds, we observed a trend suggesting that the perturbations were easier to recover from over the length of the trial.

Moving forward, adding the responses from older adults, children, and persons with lower-extremity asymmetry (i.e. stroke survivors, persons with amputations) to this protocol would provide additional insight into the influence of age, walking confidence, and inherent asymmetries on proactive modifications to stability, especially if these participants are unable to rely on their reactive capabilities to recover from perturbations. Electromyography recordings of lower extremity muscles will also provide valuable understanding as to how participants use neuromuscular control to implement these proactive modifications to stability. As we better understand how stability is proactively modified to maintain stability, rehabilitation protocols can be strengthened to target those at risk for falls. Many reactive balance interventions focus on exercise, perturbations while standing, or the post-perturbation response while walking [32,147–151]. These new results show that anteroposterior stability while walking is modifiable; therefore, dynamic stability prior to perturbations can be a target for fall-prevention interventions. The extent to which clinical populations can modify stability and the ability to transfer these modifications to walking without the threat of perturbations is still unknown.

2.5 Conclusion

These results indicate that beneficial modifications to posterior MoS at foot strike are indeed possible in an unimpaired population. At mid-swing, there was a detrimental increase in anterior instability only during stance on the dominant limb for trials with the threat of anterior perturbations compared to no perturbations. During trials with perturbations, participants decreased step lengths and increased step rates. At the slow walking speed, participants decreased the time spent in double support with the threat of posterior perturbations. These proactive modifications to stability were implemented despite the capacity for unimpaired participants to rely on their ability to recover from perturbations. Consequently, anteroposterior stability may be a feasible target for fall-prevention interventions by targeting modifications in step lengths or step rates while maintaining the same walking speed. These results also provide a framework with which to interpret results from populations with impairments.

Chapter 3

WALKING STABILITY AND LATERALITY ACROSS BALANCE DOMAINS

3.1 Introduction

Laterality, defined here as the "existence of limb dominance" [85] or the presence of asymmetry rather than symmetry, has been detailed previously in the upper extremities, showing between-arm differences in preference, specialization, or adaptation for some tasks [86–95]. Laterality may reflect limb-specific specialization, or "dynamic dominance", with the dominant limb prioritizing predictive control (i.e. developing accurate movement patterns to execute tasks) and the non-dominant limb prioritizing impedance control (i.e. incorporating feedback to adjust movement patterns to reduce errors) [88,92–96]. Laterality allows for more specific optimization and precision of movements than non-specialized movement organization, thus facilitating a greater range and depth of proficient movements [90,92,93]. The findings of these studies in upper-extremity functions invite a better description of lower-extremity function than the common single-limb dominance approach.

Symmetry of gait variables in unimpaired people has previously been assumed across a wide range of analyses [78–84]. In our previous study of walking stability in typically developing children [41], we observed an unexpected asymmetry such that the "dominant limb" had more lateral stability (Figure 3.1).

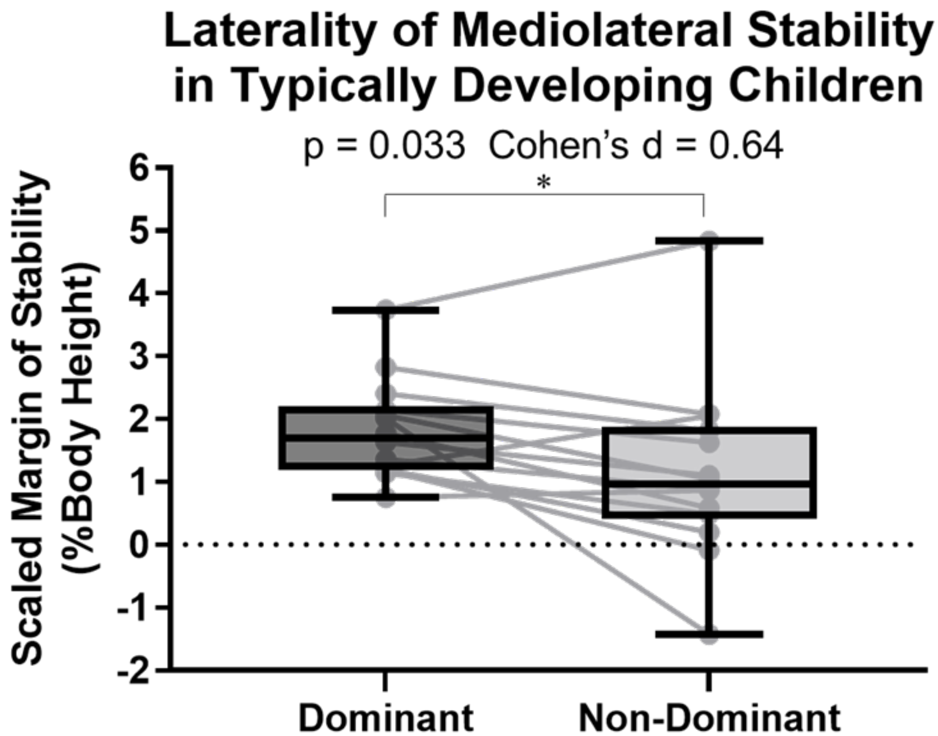

Figure 3.1: **Asymmetry between limbs for minimum lateral margin of stability for children with typical development**. Data related to a previous publication by Tracy and colleagues [41]. Limb dominance was determined by preferred kicking limb. Participants were significantly less stable in the frontal plane when supporting weight with the non-dominant limb. Limb median and 5 to 95 percentiles shown with box plots. Individual participant data are shown with light gray circles.

During gait, one limb may prioritize stability while the other limb prioritizes mobility [85,97]. Under this hypothesized specialization of limbs, the stability limb would provide a greater contribution to body support while the mobility limb would contribute more to propulsion and movement trajectories. Other studies have investigated this functional asymmetry finding evidence to support [84,85,97–100] and/or refute [97–103] this theory of lower-limb laterality in gait kinetics and kinematics, but have not pursued measures related to stability. If such laterality exists, perhaps the stability and mobility priorities respectively coincide with the impedance and predictive control model hypothesized for the upper extremities. If limb laterality underlies the asymmetry with which we maintain stability during gait, then we would expect that degree of laterality to persist across tasks that challenge stability (impedance control) and mobility (predictive control). Four proposed “domains” of balance (Figure 3.2) represent a spectrum of quasi-static to dynamic tasks, including gait, that alter the priorities of stability and mobility. Task performance across domains may not be strongly correlated [104–107], likely due to the different demands of mobility and stability. However, we anticipated that the degree of laterality, which may reflect the stability/mobility priorities of the limb, would persist across these tasks of differing demands.

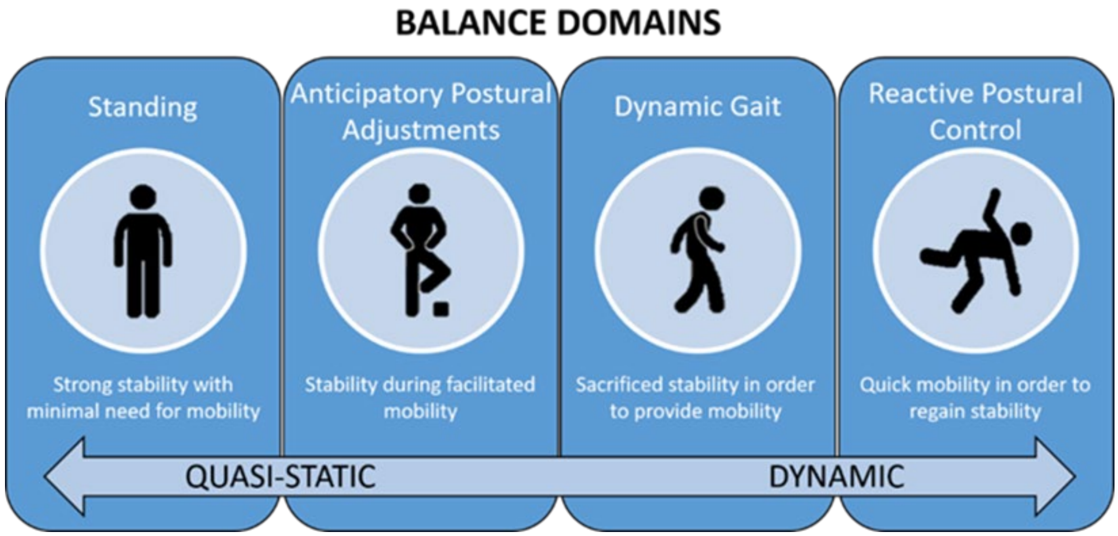

Figure 3.2: **Representation of the four primary balance domains.** Four primary balance domains relevant to postural control have been identified representing a spectrum of quasi-static and dynamic tasks: standing, anticipatory postural adjustments, dynamic gait, and reactive postural control [108–110]. Standing – maintaining control of the whole-body center of mass within the base of support while standing. This control is done by making small, continuous adjustments that control the center of mass to counteract the destabilizing force of gravity [110,111]. Anticipatory Postural Adjustments – postural control while self-initiating movements. In preparation for movement, predictive adjustments are made that will maintain stability and support mobility [106,110–112]. An example of this control is in the shifting of weight while standing towards one limb in preparation to take a step with the contralateral limb. Dynamic Gait – controlling stability and mobility while in motion. This complex coordination involves implementing proactive adjustments to impending perturbations (i.e. feedforward control) and modifying those responses to the environment (i.e. feedback control) [106,110,111]. Reactive Postural Control – coordinated whole-body responses to external perturbations. To prevent a fall, this requires rapid responses to perturbations of initially unknown magnitude or direction and can be completed with feet-in-place or compensatory stepping responses [110,111].

Beyond the common definitions of “dominant” and “non-dominant” limbs, these functional definitions of limb laterality would provide a more informed approach to rehabilitation from injuries or impairments that exaggerate asymmetry, present with asymmetrical function, and/or have a higher risk of falling (i.e. stroke survivors, persons with cerebral palsy, lower-limb amputees, leg-length discrepancy, post-ACL reconstruction surgery, and persons with Parkinson's disease) [41,115–126]. The purpose of this study was to establish the relationships between the laterality of gait stability and lower-extremity function across balance domains. We hypothesized that asymmetry in frontal-plane walking stability would relate to asymmetry during tasks in other balance domains. With this framework, we predicted that unimpaired participants would exhibit significant correlations between walking stability asymmetry and other balance domain task asymmetries. We also conducted an exploratory analysis comparing biomechanical outcomes between preferred and non-preferred stepping for gait initiation, simulated trips, and simulated slips. We hypothesized that if there are specific limb roles, that changing roles would alter performance or mechanics. We predicted that task performance would be altered when stepping with the non-preferred limb.

3.2 Methods

3.2.1 Participants

The University of Delaware IRB approved this study (Appendix E), and 30 unimpaired young adults, ages 18-40 years, provided informed consent to participate. Participants included a convenience sample of 15 females and 15 males who had no self-reported neurological or musculoskeletal disorders; no recent neural, muscular, or skeletal injuries; and no movement impairments. Additional participant descriptions included in Appendix B (Table B.1). A summary description of the participant's athletic/activity history, based on a survey modified from the Compendium of Physical Activities – Updated [152] and the CARE Consortium Baseline Questionnaire [153], is reported in Appendix B (Table B.2).

Table 3.1: **Description of participants.**

Descriptor	Sex	n	Mean	SD
Age (years)	Females	15	24.7	3.7
	Males	15	26.9	6.4
	All	30	25.8	5.2
Height (cm)	Females	15	169.8	6.7
	Males	15	183.4	7.6
	All	30	176.6	9.9
Mass (kg)	Females	15	63.0	9.6
	Males	15	75.7	12.2
	All	30	69.4	12.6
Body Mass Index ($kg \cdot m^{-2}$)	Females	15	21.9	3.3
	Males	15	22.5	3.2
	All	30	22.2	3.2

3.2.2 Protocol

Study participation included one visit to the KAAP Biomechanics Laboratory at the University of Delaware STAR Health Sciences Complex. After receiving informed consent and ensuring inclusion/exclusion criteria were met, we recorded the participant's sex and age and anthropometric data including height, mass, foot length [154], foot width [154], and leg length [124].

Participants then completed the series of balance and mobility tasks described in Table 3.2 that included tasks from each balance domain (Figure 3.3). During all testing, the participant's movement was recorded with motion capture technology (Qualisys, Gothenburg, Sweden, 120 Hz) using a 41-point whole-body reflective marker set and, when applicable, two forces plates (AMTI, MA, USA; 1200 Hz). For the computer-controlled treadmill perturbations (ActiveStep®, Simbex, Lebanon, NH, USA), participants were equipped with a safety harness system attached to an overhead rail to arrest a fall, should any occur, prior to the knees or hands touching the treadmill or floor. This system also had an in-series strain gauge (Dillon, Fairmont, MN, USA, 100 Hz) to measure the support received by a participant. Harness support was classified as a failed recovery or fall if the participant measured a peak force surpassing 20% of their body weight [133]. If the participant fell into the harness, the session was paused until the participant returned to a standing position on the treadmill and was ready to continue.

Table 3.2: **Description of balance and mobility tasks.**

Task	Domain	Description
Overground Walking [41]	Gait	Participant walked at a continuous, self-selected speed for 10 m across a ground-level surface for 6-8 trials.
Standing Postural Sway [36,104]	Standing	Participant stood with each foot on a separate force plate and were instructed to stand as still as possible. Sway was quantified for eyes-closed and eyes-opened 30-s trials.
Gait Initiation [155–157]	Anticipatory	Starting from a standing position, participant walked at a self-selected speed for 6 m, repeated for 6 trials. If one limb was not self-selected for the initial step, the participant completed two trials prompted to step with the opposite foot.
Anteroposterior Treadmill Perturbations [29,30]	Reactive	Participant received forward and backward treadmill-induced standing perturbations, six each, in a pseudorandomized order. The direction was unknown to the participant, and the size of the perturbations were sufficiently large enough to evoke a protective step. If one limb was not self-selected for the initial step, the participant completed two trials prompted to step with the opposite foot.

Note: Tasks were included from each of the four balance domains: standing, anticipatory postural control, dynamic gait, and reactive postural control [108–110]. Inter-limb asymmetry from tasks were compared to the asymmetry in walking stability from the overground walking task.

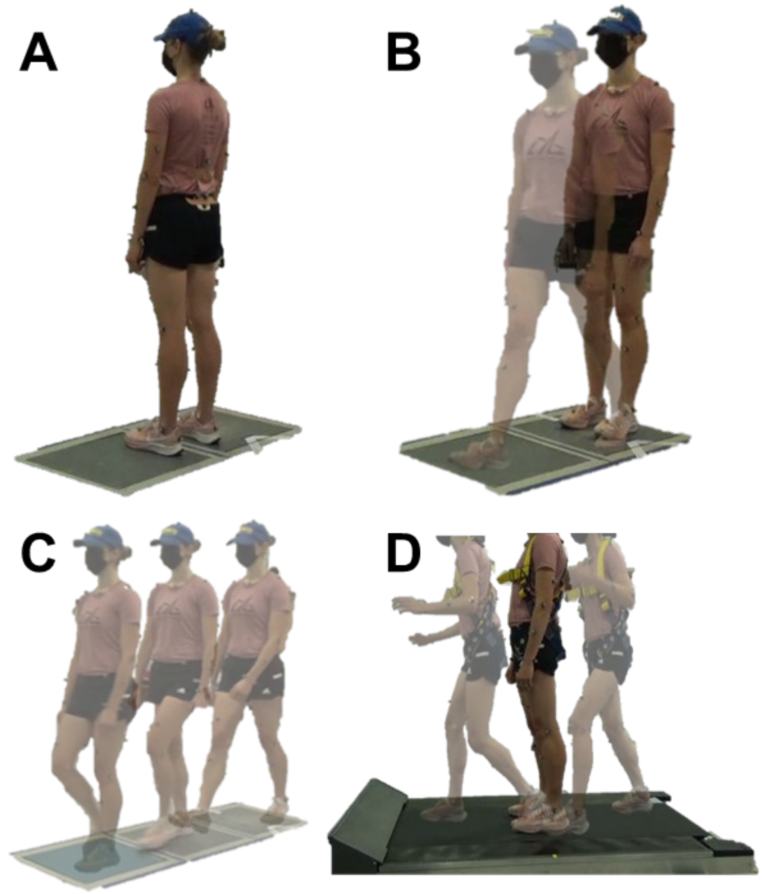

Figure 3.3: **Illustration of the four balance and mobility tasks.** Panel A – Standing postural sway (standing balance domain). Panel B – Gait initiation (anticipatory postural control balance domain). Panel C – Dynamic gait (gait balance domain). Panel D – Treadmill perturbations (reactive postural control balance domain). *Images used with participant's permission.

To encourage unbiased responses, participants were not initially informed about the study aims regarding laterality, rather, that the study was generally about balance and mobility. If the participant asked which limb to use, we instructed the participant to choose either limb or use whichever limb naturally reacts to performing the task. At the conclusion of the session, participants completed the Waterloo

Footedness Questionnaire – Revised [128] (Appendix A Figure A.1) to characterize limb dominance and were informed of the purpose of the study regarding limb choice and laterality.

3.2.3 Analysis

3.2.3.1 Inter-Limb Asymmetry

We quantified the degree to which participants were asymmetric between limbs using the Inter-Limb Asymmetry (ILA, Equation 3.1).

$$ILA = \left(\frac{Left\ Limb - Right\ Limb}{scaling\ factor}\right) \cdot 100 \qquad \text{Equation 3.1}$$

The ILA measures the between-limb difference and scales that value to the participant's height. A negative percentage indicates a greater value on the right limb, while a positive percentage indicates a greater value on the left limb. The ILA measure is modified from Carpes and colleagues [114], and is explained in more detail in Appendix C.

3.2.3.2 Balance and Mobility Tasks

As the primary focus of this study was to understand the underlying mechanisms of gait stability asymmetry, the measure from overground walking was the reference to which all other asymmetry measures were compared (Gait Balance Domain, Table 3.2, Figure 3.3C). Two consecutive gait cycles for each limb were evaluated from the middle of each walking trial. The minimum lateral MoS (Figure 3.4) [35,41] was calculated using custom LabVIEW software (National Instruments, Austin, TX, USA, 2018) and scaled to the participant's height [41,138]. Within limb values were averaged within and across trials. The ILA was calculated according to equation 3.1. The ‘Left Limb’ and ‘Right Limb’ values were the left and right minimum lateral MoS values, respectively. The difference between these two values was scaled by participant height and expressed as a percentage of height.

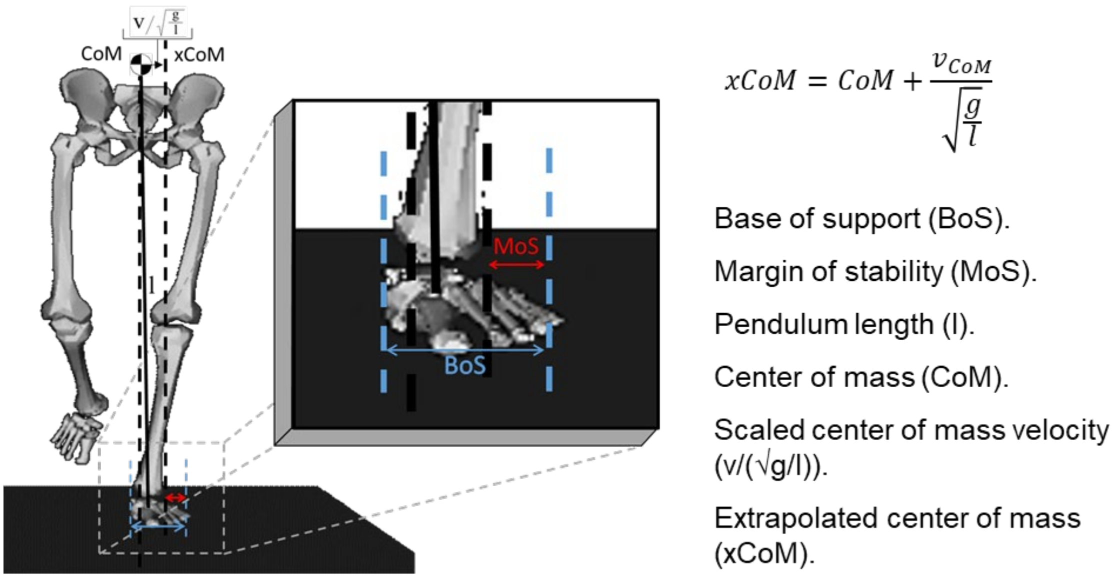

Figure 3.4: **Illustration of the minimum lateral margin of stability.** Figure modified from previous publication [41]. The minimum lateral margin of stability represents the smallest distance between the base of support and the extrapolated center of mass (center of mass position + scaled center of mass velocity) during a single gait cycle. Positive values represent a state of stability (i.e. the extrapolated center of mass is within the base of support or advantageously placed away from the edge of the base of support), and a perturbation is needed to initiate a fall in that direction. Negative values represent a state of instability (i.e. the extrapolated center of mass is outside the base of support), and a compensatory action such as taking a step, applying an external force, or counter-rotating segments about the center of mass is needed to prevent a fall.

For the standing balance task (Standing Balance Domain, Table 3.2, Figure 3.3A), the net center of pressure (COP) trajectory was calculated as the weighted sum of the time-varying position of the COP from each force plate [158] using a custom Excel Visual Basic for Applications script (Microsoft, Redmond, WA, USA, version 2016). We focused on the anteroposterior (AP) component of the COP as it is a better indicator of sway control using the ankle strategy—the primary strategy employed during unperturbed stance [36]. The mediolateral component of COP is more influenced by shear forces and the distribution of body weight between limbs utilizing the hip strategy [36]. The relationship between the net AP COP and the AP COP for each limb was evaluated with linear regression (Equation 3.2):

$$COP_{net} = \alpha + \beta(COP_{limb}), \qquad \text{Equation 3.2}$$

where COP_{limb} represents either the left or right limb AP COP trajectory, and β represents the slope of the linear regression. The ILA value was calculated as the magnitude of difference between the left and right limb beta values and expressed as a percentage. This deviates from equation 3.1 in that this value did not need to be scaled to the participant.

A beta value of 1.0 indicates a 1:1 ratio of COP_{net} and COP_{limb} trajectories. A beta value with a magnitude less than one indicates that a given displacement in COP_{limb} corresponds with a smaller displacement of COP_{net}, and a beta value with a magnitude greater than one with a greater displacement of COP_{net} for the COP_{limb} displacement. Using a novel approach, we interpreted $\beta < 1.0$ as evidence of active limb control of anteroposterior sway and $\beta > 1.0$ of passive limb control. Proof-of-concept trials with and without restricted COP capabilities beneath one limb supported this interpretation (Figure 3.5).

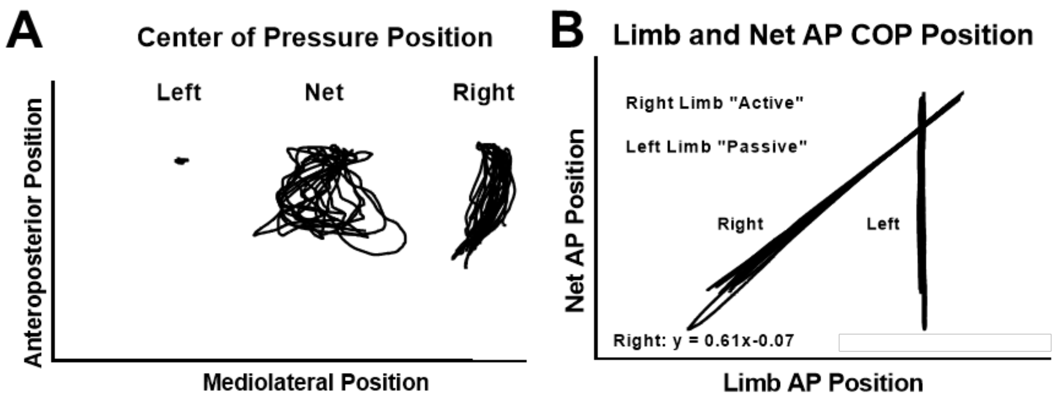

Figure 3.5: **Example proof-of-concept trial for beta coefficient interpretation**. Panel A – Individual limb and net center of pressure trajectories. The left limb was placed on a small object limiting the center of pressure motion to a small space. Panel B – Individual limb regressions to net center of pressure. We interpreted $\beta < 1$ as evidence of active limb control of anteroposterior sway and $\beta > 1$ of passive limb control.

For the gait initiation task (Anticipatory Balance Domain, Table 3.2, Figure 3.3B), the anticipatory postural adjustment (APA) phase represents the initial shifting of the body to generate lateral and propulsive forces for movement (Figure 3.6). Using a custom Excel Visual Basic for Applications script (Microsoft, Redmond, WA, USA, version 2016), the beginning of the APA phase (i.e. movement initiation) was marked at the point where the vertical force moved beyond a 2 standard deviation threshold of the mean vertical force across the first 0.5 s during standing, and the end of the APA phase was marked at the point when the center of pressure was closest to the stepping limb [157]. The ILA value was calculated according to equation 3.1. The 'Left Limb' value was the mediolateral displacement of the center of pressure during the APA phase when stepping with the left limb. The 'Right Limb' value was the mediolateral displacement of the center of pressure during the APA phase when stepping with the right limb. The difference between these two values was scaled by the participant's height and expressed as a percentage of height. The primary purpose of the APA phase of gait initiation is to shift the center of pressure to a position that will move the center of mass from a centrally-located position towards the initial stance limb allowing the stepping limb to be unweighted [157,159]. We focused on the mediolateral component of the APA phase of gait initiation because it represents this initial anticipatory shifting of the center of pressure that will move the center of mass towards the stance limb. There is also typically some forward momentum generated during this phase with a posterior displacement of the center of pressure [157,159]; however, the initial stance limb generates the majority of the propulsion needed for gait initiation during the later locomotor phase, and we wanted to focus on the anticipatory nature of this task.

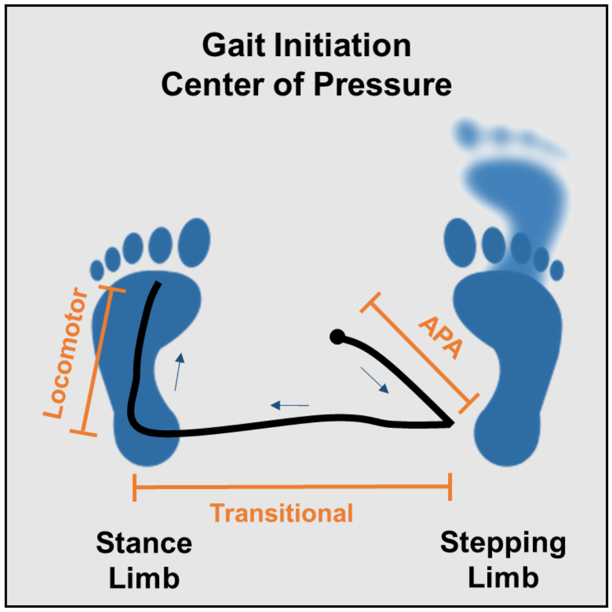

Figure 3.6: **Illustration of the center of pressure trajectory during gait initiation.** The anticipatory postural adjustment (APA) phase represents the initial shifting of the body to generate lateral and propulsive impulses. The transitional phase represents the movement of the center of pressure to the initial stance limb to unweight the stepping limb. The locomotor phase represents the forward progression of the center of pressure during the step.

For the treadmill perturbation task (Reactive Balance Domain, Table 3.2, Figure 3.3D), perturbations consisted of triangular-shaped velocity profiles with a 0.4 s duration (ActiveStep®, Simbex, Lebanon, NH, USA). Anterior perturbations (i.e. simulated trips with 0.6 m displacement) reached a peak velocity of 1.5 $m \cdot s^{-1}$ (acceleration of 7.5 $m \cdot s^{-2}$), and posterior perturbations (i.e. simulated slips with -0.48 m displacement) reached a peak velocity of -1.2 $m \cdot s^{-1}$ (acceleration of -6.0 $m \cdot s^{-2}$). At foot strike of the recovery step, the anteroposterior and mediolateral distances between the center of mass and the stepping limb's toe marker (anterior edge of the base of support for forward stepping) or heel marker (posterior edge of the base of support for backward stepping) were calculated using custom LabVIEW software (National Instruments, Austin, TX, USA, 2018). These measures have been identified previously as important aspects of the stepping response in fall prevention strategies [150,160,161]. The ILA values were calculated according to equation 3.1. The 'Left Limb' values were the respective anteroposterior or mediolateral distances between the center of mass and the stepping limb when stepping with the left limb. The 'right Limb' values were the respective distances when stepping with the right limb. The differences between these respective distances were scaled by the participant's height and expressed as a percentage of height.

In alignment with our purpose to establish the relationships between the laterality of gait stability and lower extremity function across balance domains, the between-limb walking stability asymmetry was correlated to the between-limb asymmetries within each task using Pearson correlations. Significance was set at $p < 0.05$. Thirty participants provided 80 percent power to detect a correlation coefficient of 0.361 or larger as significant. An additional analysis was also considered exploring the biomechanical outcomes of task performance when the limb roles were switched, when possible (e.g. non-preferred limb gait initiation and reactive stepping). These outcomes were evaluated using paired t-tests and Cohen's d effect size for repeated measures.

3.3 Results

3.3.1 Limb Dominance and Footedness

All 30 participants reported right limb dominance according to their preferred kicking limb. The average Waterloo Footedness score, which ranges from a possible score of -2 (completely left-footed) to +2 (completely right-footed), was 0.55 (SD: 0.46; Range: -0.30 to 1.60, Figure 3.7, Appendix A Figure A.1).

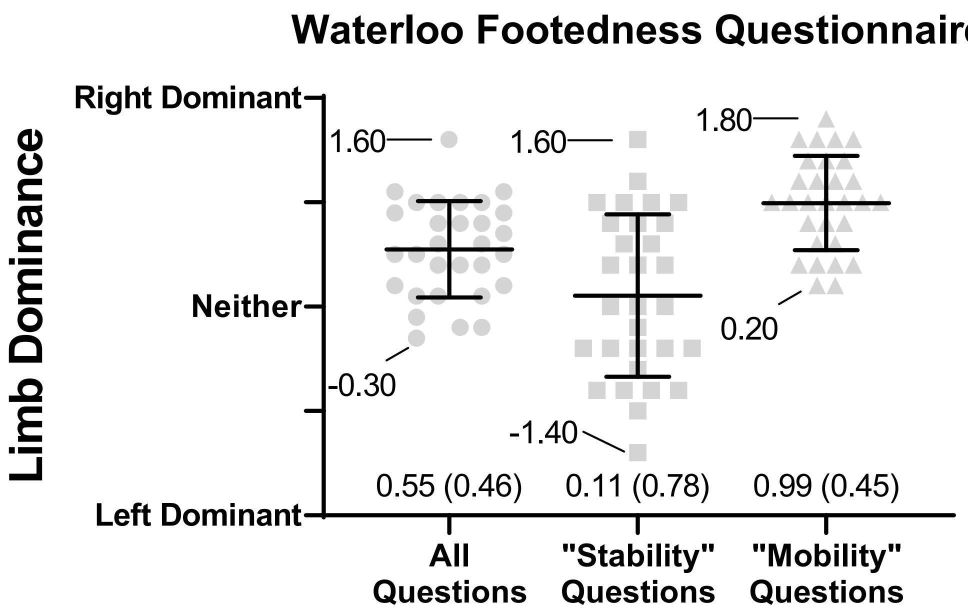

Figure 3.7: **Results from the Waterloo Footedness Questionnaire—Revised** [128]. Participants responded to ten questions, five related to "stability" tasks and five related to "mobility" tasks, with which limb they would perform a task (left, right, or equal) and the extent to which they would use that limb for the task (always or usually). See also Appendix A Figure A.1.

3.3.2 Inter-Limb Asymmetry Correlations

There were no significant correlations between the ILA for minimum lateral MoS during walking and the respective ILA values for each task ($r < 0.327$, $p > 0.119$, Figure 3.8). The ILA values showed no clear dominance across participants (Figure 3.8). The means, standard deviations, minimums, and maximums of ILA values for each task are presented in Appendix B (Table B.3). The left and right limb task measurement means, standard deviations, minimums, and maximums are presented in Appendix B (Table B.4).

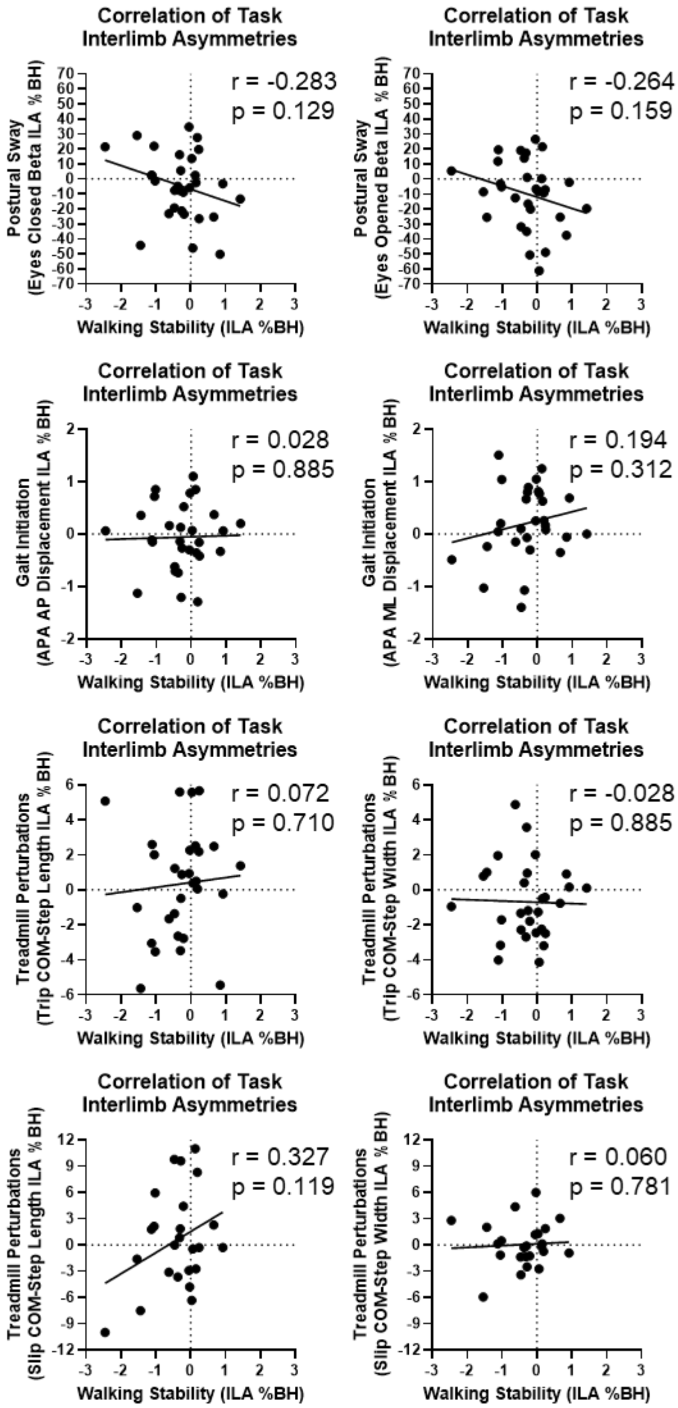
Correlation of Task Interlimb Asymmetries
Postural Sway (Eyes Closed Beta ILA % BH)
r = -0.283
p = 0.129
Walking Stability (ILA %BH)
Correlation of Task Interlimb Asymmetries
Postural Sway (Eyes Opened Beta ILA % BH)
r = -0.264
p = 0.159
Walking Stability (ILA %BH)
Correlation of Task Interlimb Asymmetries
Gait Initiation (APA AP Displacement ILA % BH)
r = 0.028
p = 0.885
Walking Stability (ILA %BH)
Correlation of Task Interlimb Asymmetries
Gait Initiation (APA ML Displacement ILA % BH)
r = 0.194
p = 0.312
Walking Stability (ILA %BH)
Correlation of Task Interlimb Asymmetries
Treadmill Perturbations (Trip COM-Step Length ILA % BH)
r = 0.072
p = 0.710
Walking Stability (ILA %BH)
Correlation of Task Interlimb Asymmetries
Treadmill Perturbations (Trip COM-Step Width ILA % BH)
r = -0.028
p = 0.885
Walking Stability (ILA %BH)
Correlation of Task Interlimb Asymmetries
Treadmill Perturbations (Slip COM-Step Length ILA % BH)
r = 0.327
p = 0.119
Walking Stability (ILA %BH)
Correlation of Task Interlimb Asymmetries
Treadmill Perturbations (Slip COM-Step Width ILA % BH)
r = 0.060
p = 0.781
Walking Stability (ILA %BH)

Figure 3.8: **Correlation of lateral walking stability and the other task inter-limb asymmetries.** The ILA measures the between-limb difference and scales that value to the participant's height. A negative percentage indicates a greater value on the right limb, while a positive percentage indicates a greater value on the left limb. There were no significant correlations between the ILA value for minimum lateral MoS during walking and the respective ILA values for each task in the other balance domains.

3.3.3 Stepping Tasks with Preferred and Non-Preferred Limbs

Stepping tasks were completed with the self-selected limb for each trial. Preferred and non-preferred stepping limbs were determined by the majority choice for six trials, and no limb preference was determined if both limbs were self-selected equally. The preferred and non-preferred limb stepping task measurement means, standard deviations, minimums, and maximums are presented in Appendix B (Table B.5).

When initiating gait with the non-preferred limb, the mediolateral displacement of the center of pressure during the APA phase was significantly greater ($p = 0.04$, $d = 0.43$, mean difference (SD) 0.47 (1.09) cm) and the posterior displacement of the center of pressure was not different ($p = 0.99$, $d < 0.01$, < 0.01 (1.16) cm) compared to initiating gait with the preferred limb (Figure 3.9).

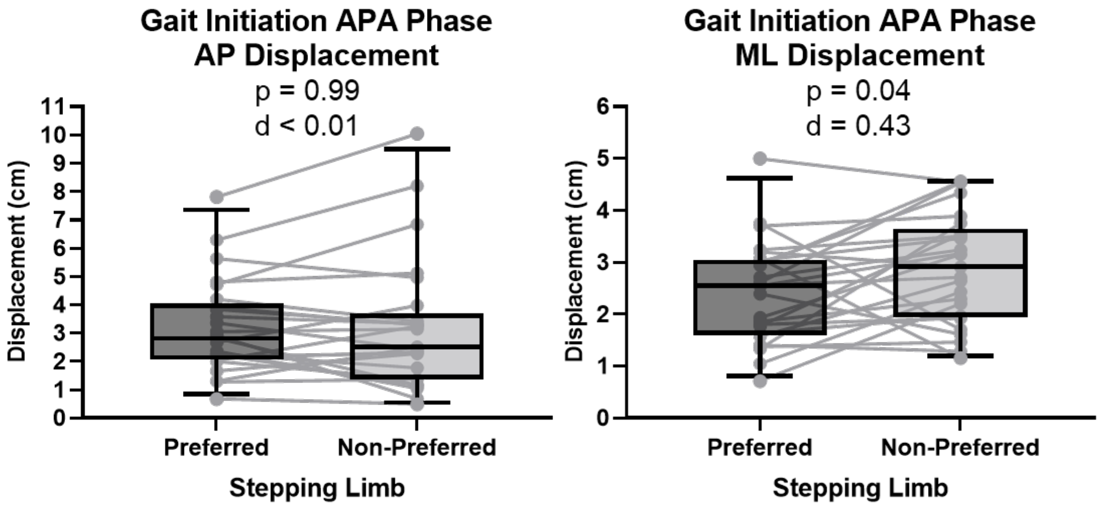

Figure 3.9: **Gait initiation center of pressure displacement when stepping with the preferred and non-preferred limbs.** When stepping with the non-preferred limb, the COP mediolateral displacement of the anticipatory postural adjustment (APA) phase was significantly greater and the anteroposterior displacement was no different compared to initiating gait with the preferred limb. Limb median and 5 to 95 percentiles shown with box plots. Individual participant data are shown with light gray circles.

When recovering from simulated trips with the non-preferred limb, participants had a larger anterior distance ($p = 0.04$, $d = 0.40$, mean difference (SD) 2.07 (5.12) cm, Figure 3.10) between the center of mass and anterior edge of the base of support with the center of mass posterior to the edge of the base of support.

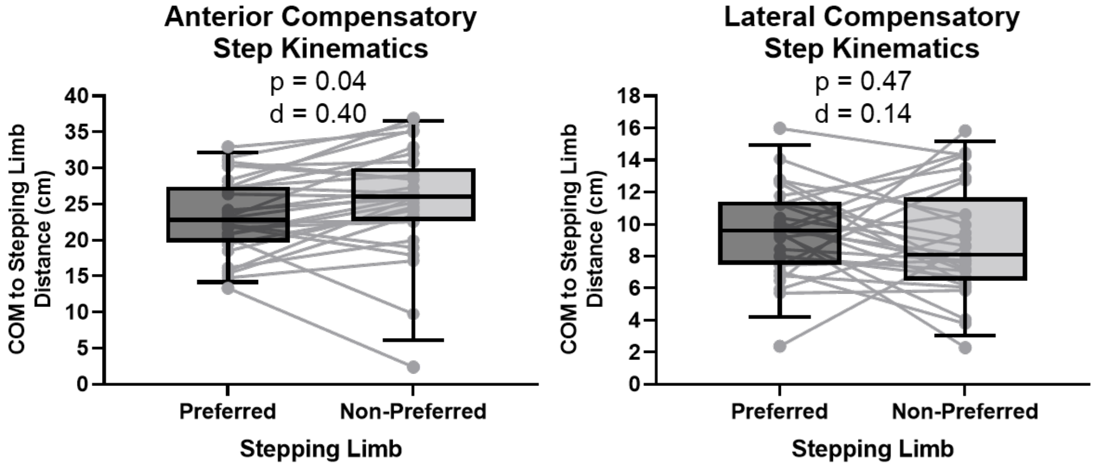

Figure 3.10: **Anterior compensatory step kinematics with the preferred and non-preferred limbs.** When stepping with the non-preferred limb, the anterior distance between the center of mass and stepping limb toe was significantly greater and the lateral distance was no different compared to taking a recovery step with the preferred limb. Limb median and 5 to 95 percentiles shown with box plots. Individual participant data are shown with light gray circles.

When recovering from simulated slips with the non-preferred limb, participants had a larger posterior distance (p = 0.04, d = 0.45, mean difference (SD) 4.07 (9.14) cm) and a smaller lateral distance (p = 0.01, d = 0.65, 2.55 (3.90) cm) between the center of mass and the posterior edge of the base of support with the center of mass anterior and medial to the edge of the base of support (Figure 3.11).

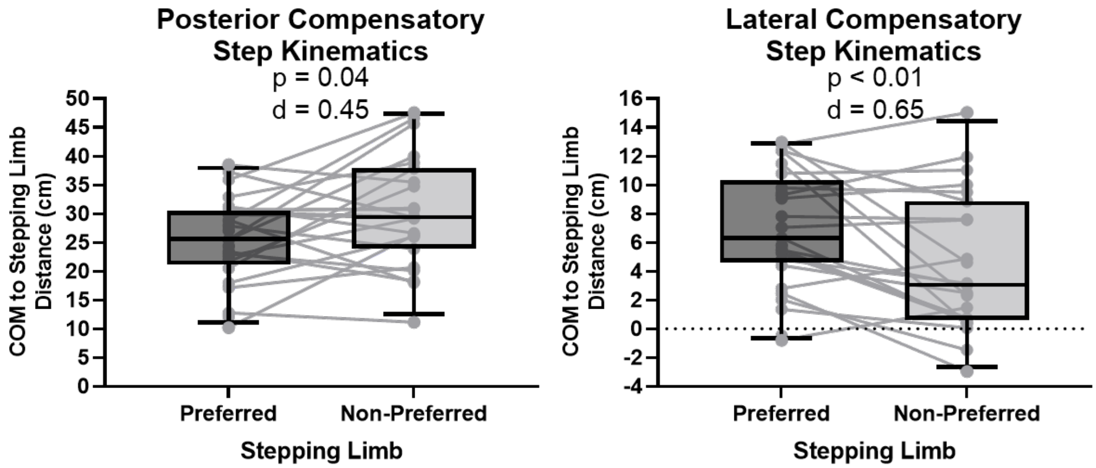

Figure 3.11: **Posterior compensatory step kinematics with the preferred and non-preferred limbs.** When stepping with the non-preferred limb, the posterior distance between the center of mass and stepping limb heel was significantly greater and the lateral distance was significantly smaller compared to taking a recovery step with the preferred limb. Limb median and 5 to 95 percentiles shown with box plots. Individual participant data are shown with light gray circles.

3.4 Discussion

3.4.1 Inter-Limb Asymmetry Correlations

The purpose of this analysis was to establish the relationships between the laterality of gait stability and lower-extremity function across balance domains. We hypothesized that asymmetry in frontal-plane walking stability would relate to asymmetry during tasks in other balance domains. With this framework, we predicted that unimpaired participants would exhibit significant correlations between walking stability asymmetry and other balance domain task asymmetries. This prediction was not supported, as there were no significant correlations observed between the ILA for walking stability and the ILA values for each task in the other balance domains (Figure 3.8). This laterality framework may exist if different tasks were compared across balance domains, exist only across tasks within a balance domain, or may not exist at all. Lower-extremity laterality may be task or balance domain specific rather than a representation of systemic lower-extremity impedance/predictive control. This result extends the evidence for the independence of balance domain task performance that we've observed previously [104,105] to include independent performance for stepping tasks.

3.4.2 Stepping Tasks with Preferred and Non-Preferred Limbs

We also conducted an exploratory analysis comparing biomechanical outcomes between preferred and non-preferred stepping for gait initiation, simulated trips, and simulated slips. We hypothesized that, if there are specific limb roles, changing those roles would alter performance or mechanics. As predicted, there were significant between-limb differences when stepping tasks (i.e. gait initiation, simulated trips, and simulated slips) were completed with the non-preferred limb.

The APA phase of gait initiation represents the initial shifting of weight towards the stepping limb to generate the impulse to shift the center of mass over to the stance limb [155–157]. During non-preferred stepping during gait initiation, there was an increase in the mediolateral displacement of the COP (Figure 3.9). It has been suggested that a greater mediolateral displacement of the COP during the APA phase for patients with Parkinson's disease could be evidence of a decoupling of forward momentum generation between the stance and stepping limbs [159,162]. When our participants step with their non-preferred limb, we may be seeing a similar separation where the stepping limb increases its unique contribution to lateral momentum. The greater displacement in the mediolateral component of the APA phase during non-preferred stepping may not have a functional difference, though. There were no statistical differences in mediolateral step width (p = 0.18, d = 0.28) or mediolateral step velocity (p = 0.15, d = 0.30) during non-preferred stepping. Previously, stepping with the non-preferred limb has resulted in a wider step [163]. There was a meaningful methodological difference in this study, though, as participants began walking in response to a visual cue in contrast to this study where participants self-selected a start time. This changes the task's balance domain from an anticipatory to a reactive balance task, and the results in this chapter support the independence of balance domains shown previously [104,105]. With repetition, we expect that the increased familiarity with stepping with the non-preferred limb would decrease differences between stepping limb performances. During non-preferred stepping, there was no difference in the anteroposterior displacement of the COP during the APA phase (Figure 3.9). This comparable result suggests that both limbs performed similarly regarding the generation of anteriorly directed forces as part of the initial propulsive impulse. Accordingly, there were no statistical differences in anteroposterior step length (p = 0.35, Cohen's d = 0.20) or anteroposterior step velocity (p = 0.89, d = 0.03) between limbs.

When stepping with the non-preferred limb during treadmill perturbations, participants had a larger anterior distance between the center of mass and the stepping limb toe (i.e. the anterior edge of the base of support) during simulated trips (Figure 3.10) and a larger posterior and smaller lateral distance between the center of mass and the stepping limb heel (i.e. the posterior edge of the base of support) during simulated slips (Figure 3.11). These adjustments likely represent a compensation towards increasing stability. Individuals with chronic stroke increased their step lengths through fall-recovery training in a manner that improved their ability to recover from simulated trips [150], larger step lengths have been associated with increased stability during slips [160], and increased lateral distance of foot placement has been associated with increased fall risk during slips [161].When stepping with the non-preferred limb, participants may have decreased confidence in preventing a fall and consequentially increased their response towards stability to improve their ability to recover. Switching the roles between preferred and non-preferred stepping limbs also changes which limb functions as the stance limb. The stance limb has been shown to contribute to trip recoveries by quickly working (latencies < 100 ms) to arrest angular momentum and increase the time for proper positioning of the recovery limb [49,164,165]. Therefore, altering which limb utilizes the stance limb contribution to recovery may decrease the impact that the stance limb can have on recovery requiring a greater contribution of the stepping limb. With repetition, we expect that the increased familiarity with stepping with the non-preferred limb would decrease differences between stepping limb performances.

3.4.3 Stepping Limb Preferences across Tasks and Limb Dominance

Although not a planned analysis of this study, during our comparisons of preferred and non-preferred stepping we noted that stepping limb preferences were not consistent within or between tasks and were discordant with self-reported limb dominance (Table 3.3). Participants did tend to have preferred stepping limbs for gait initiation and treadmill perturbations (i.e. tasks that could be completed with either limb), but limb dominance or stepping limb preference during the other stepping tasks may not predict that limb preference. The preferred stepping limbs were largely mixed between the dominant and non-dominant limbs within tasks and between stepping tasks (Table 3.3). All participants were able to complete the gait initiation task with the non-preferred stepping limb either spontaneously or when instructed; however, even when instructed, one participant was unable to step with the non-preferred limb during simulated trips, and six participants were unable to step with the non-preferred limb during simulated slips. Future studies should consider participant preferences for performing stepping tasks within their methods and analyses, as self-reported limb dominance may have little relevance to task performance preference, and task execution may change when completed with the non-preferred limb.

Table 3.3: **Self-reported limb dominance and stepping task limb preferences**.

			Self-Reported Limb Dominance			Gait Initiation Stepping Limb			Simulated Trips Stepping Limb		
			L	N	R	L	N	R	L	N	R
Individual Tasks	Simulated Slips	L	0	0	17	5	2	10	2	0	15
		N	0	0	1	0	0	1	1	0	0
		R	0	0	12	4	2	6	8	0	4
	Simulated Trips	L	0	0	19	4	2	13			
		N	0	0	0	0	0	0			
		R	0	0	11	5	2	4			
	Gait Initiation	L	0	0	17						
		N	0	0	4						
		R	0	0	9						
Stepping Tasks	Preferred Stepping Limb (all)	L	0	0	10						
		M	0	0	17						
		R	0	0	3						

Note: All 30 participants self-reported right limb dominance as determined by the limb used to kick a ball. Individual Tasks – We determined the stepping limb preference by which limb was used for the majority of six stepping responses for that task. A left stepping preference indicated with 'L' row or column, no preference indicated with 'N', and a right preference indicated with 'R'. Stepping Tasks – The stepping limb preference combined for all three stepping tasks (i.e. gait initiation, simulated trips, and simulated slips). A left stepping preference for all tasks indicated with 'L' row, a mixed preference indicated with 'M', and a right preference indicated with 'R'.

All 30 of our participants self-reported as right limb dominant according to preferred kicking limb (Table 3.3). It is unclear if left-limb dominant participants would be consistent with these results. From a previous report of how lower-limb dominance was distributed across 3,307 healthy adults using the Lateral Preference Inventory, mixed-footedness was quite common (female: 47.3%; male: 59.4%), and truly left-footed participants were quite rare (female: 2.2%; male: 1.9%) [129]. Within our own participant group using the Waterloo Footedness Questionnaire [128], none of the 30 participants were completely right- or left-footed as a composite score or within the subsets of "stability" and "mobility" questions (Figure 3.7, see also Appendix A Figure A.1). Therefore, lower-extremity frameworks that operate with a dichotomized left- or right-limb dominance approach may be too simple and may not account for balance domain or task-specific stability/mobility needs.

3.4.4 Walking Stability Asymmetry

This study was motivated, in part, by previously observed, between-limb asymmetry in lateral walking stability of typically developing children ($p = 0.033$, Cohen's $d = 0.64$, Figure 3.1 and Figure 3.12) [41]. Such asymmetry was not replicated with this sample of unimpaired adults ($p = 0.062$, $d = 0.35$, Figure 3.12). However, the anticipated between-limb asymmetry was observed within the similar participant group presented in Chapter 2 walking at an estimated preferred speed on a treadmill without perturbations ($p = 0.001$, $d = 1.23$, Figure 3.12). The participants presented in Chapter 2 and in this study were of similar age and functionality groups, but completed walking with different modes (i.e. treadmill vs. over ground), so age does not seem to be the determining factor. The typically developing children and the participants in this analysis both completed walking with the same mode (i.e. over ground), so walking mode does not seem to be the determining factor either. It is unclear as to why this between-limb asymmetry in lateral stability was seen in these two previous settings but not this current one. Perhaps there exists an interaction between age and walking surface where the young adult participants do not have the between-limb asymmetry in lateral walking stability shown in the typically developing children group when walking over ground, but the narrow width of the treadmill induces a greater asymmetry in adults during treadmill walking. All three participant groups showed a similar trend of asymmetry where stance on the dominant limb was more stable (Figure 3.12). Therefore, it is reasonable to expect a small-to-moderate effect of between-limb asymmetry in lateral walking stability, even in an unimpaired population. Future studies should not assume symmetry between limbs for lateral stability for over ground or treadmill walking.

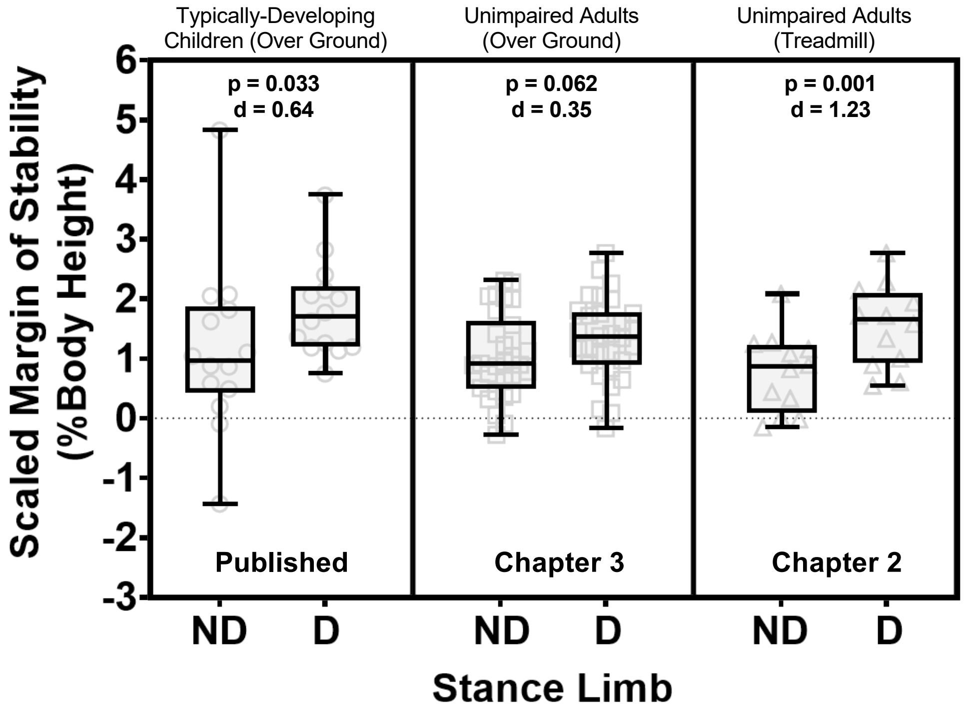

Figure 3.12: **Between-limb asymmetry trend in lateral stability**. A significant between-limb asymmetry in lateral walking stability was observed in a previous study of typically developing children ($p = 0.033$, Cohen's $d = 0.64$) [41]. This asymmetry was not observed in the participants presented here in Chapter 3 ($p = 0.062$, $d = 0.35$), but was observed in the unimpaired adults presented in Chapter 2 ($p = 0.001$, $d = 1.23$).

This work was the first to look at between-limb asymmetries across balance domains under the hypothesized stability and mobility specialization of limbs framework. Stability and mobility are logical, functional targets for addressing fall risk and barriers to physical activity. As evident by the domains of balance (Figure 3.2), stability and mobility are interrelated, and often competing, in daily tasks. Limbs often function in roles where priorities are ambiguous such as walking, are switched such as stepping with the non-preferred limb, and asymmetrical such as turning. Previous studies focused on laterality found the left limb to be responsible for support (i.e.

“stability limb”) and the right limb to be responsible for propulsion (i.e. “mobility limb”) [84,85,97–100]; although, this structure has not always been supported by other results [97–103]. Our current results reveal limb preferences and differences in task performance, but self-reported limb dominance does not provide clarity regarding limb preference and limb function across balance domains (Table 3.2). Echoing Gabbard and Hart, “the basic question of foot dominance remains somewhat unsettled” [97]. The concept of limb mobility/stability priorities, and the potential relationship to impedance/predictive control, remains possible. Our results suggest that these limb roles may not transfer across balance domains (Figure 3.8). However, with so many participants showing a limb preference for stepping tasks (Table 3.3), they may be capitalizing on the benefits of laterality for increased optimization and precision of movements as seen in the upper extremities [90,92,93]. The next step is to investigate between-limb asymmetries within balance domains as limb priorities may persist across tasks with similar stability and mobility necessities. If inter-limb asymmetries do not persist within balance domains, then limb preferences and priorities may be simply task-specific.

Of future interest would be evaluating relationships between asymmetry and preferred/non-preferred task performance within groups with characteristic asymmetries. Inherent asymmetries in populations such as stroke survivors and persons with lower-limb loss likely generate a greater limb preference within movement tasks. Populations with characteristic asymmetries and an elevated risk of falling may have much greater differences in task performance when completed with the non-preferred limb. In an unimpaired population, like that of this study, the non-preferred limb still functions quite well. No participant fell into the harness during any of the simulated trip or slip trials with the preferred or non-preferred stepping limb, whereas 5 out of 14 participants with chronic stroke fell into the harness due to much smaller anterior perturbations of the same design [150]. This work also provides a framework with which to interpret asymmetries in additional populations.

3.5 Conclusion

These results showed no significant correlations between walking inter-limb asymmetry and inter-limb asymmetries from other balance domain tasks. Participants did exhibit strong tendencies towards limb preferences within tasks, but commonly used self-reported limb dominance did not seem to predict those preferences. Participants also showed between-limb differences when performing stepping tasks with the non-preferred limb. Future studies should not assume between-limb symmetry in stability or assume that self-reported limb dominance is a meaningful predictor of task limb preference or performance.

Chapter 4

SUMMARY CONCLUSIONS AND FUTURE DIRECTIONS

4.1 Aim 1 Summary – Proactive Modifications to Walking Stability

The purpose of the first aim of this dissertation, presented in Chapter 2, was to investigate the possibility of proactive modifications to stability using the threat of large anterior and posterior perturbations during walking at multiple speeds. We hypothesized that anteroposterior stability would be a modifiable aspect of gait. We predicted that unimpaired participants would increase stability protective against a loss of stability when threatened with large perturbations and would display more pronounced changes in stability at slower speeds.

With the threat of posterior perturbations, beneficial proactive modifications were observed for posterior stability at foot strike. With the threat of anterior perturbations, there was a detrimental increase in anterior instability at mid-swing only during stance on the dominant limb. Modifications of stability occurred at all walking speeds. Proactive modifications to stability were accompanied by shortened step lengths and increased step rates, but no change in step widths. Specific to walking at the slow speed and with the threat of posterior perturbations, there was also less time spent in double support.

These results indicate that beneficial modifications to stability are indeed possible in an unimpaired population. These proactive modifications to stability were implemented despite the capacity for unimpaired participants to rely on their ability to recover from perturbations. Consequently, anteroposterior stability may be a feasible target for fall-prevention interventions by targeting modifications in step lengths or step rates while maintaining the same walking speed. These results also provide a framework with which to interpret results from populations with impairments.

4.1.1 Future Directions

The extent to which clinical populations can modify stability and the ability to transfer these modifications to walking without the threat of perturbations is still unknown. Adding the responses from older adults, children, and persons with lower-extremity asymmetry (i.e. stroke survivors, persons with amputations) to this protocol would provide additional insight into the influence of age, balance confidence, and inherent asymmetries on proactive modifications to stability. Electromyography recordings of lower-extremity muscles will also provide valuable understanding as to how participants use neuromuscular control to implement these proactive modifications to stability. The relationships between modifications to stability and walking economy are also unknown. Interventions that optimize the relationship between increasing stability and minimizing effort are the most desirable targets. As we better understand how stability is proactively modified to maintain stability, rehabilitation protocols can be strengthened to target those at risk for falls. These results provide evidence that this protocol may be a feasible walking stability intervention. Additional studies are needed to evaluate if these proactive modifications can be trained. We do not know if these changes to gait are able to make a persistent change on walking mechanics in trials without perturbations or in the free-living environment.

4.2 Aim 2 Summary – Laterality across Balance Domains

The purpose of the second aim of this dissertation, presented in Chapter 3, was to establish the relationships between the laterality of gait stability and lower-extremity function across balance domains. We hypothesized that asymmetry in lateral walking stability would relate to asymmetry during tasks in other balance domains. With this framework, we predicted that unimpaired participants would exhibit significant correlations between walking stability asymmetry and other balance domain task asymmetries. We also conducted an exploratory analysis comparing biomechanical outcomes between preferred and non-preferred stepping for gait initiation, simulated trips, and simulated slips. We hypothesized that if there were specific limb roles, that changing roles would alter performance or mechanics. We predicted that task performance would be altered when stepping with the non-preferred limb.

There were no significant correlations observed between the ILA for walking stability and the ILA values for each task in the other balance domains. There were significant between-limb differences when stepping tasks (i.e. gait initiation, simulated trips, and simulated slips) were completed with the non-preferred limb. Limb preferences within stepping tasks were inconsistent with self-reported limb dominance.

These results showed no significant correlations between walking inter-limb asymmetry and tasks from other balance domains. Participants did exhibit strong tendencies towards limb preferences within tasks, but commonly used self-reported limb dominance did not seem to predict those preferences. Participants also showed between-limb differences when performing stepping tasks with the non-preferred limb. Future studies should not assume between-limb symmetry in stability or assume that self-reported limb dominance is a meaningful predictor of task limb preference or performance.

4.2.1 Future Directions

While between-limb asymmetries were not related between tasks *across* balance domains, between-limb asymmetries *within* balance domains may relate as limb priorities may persist across tasks with similar stability and mobility necessities. If inter-limb asymmetries do not persist within balance domains, then limb preferences and priorities may be simply task specific. Populations with characteristic asymmetries, such as stroke survivors and persons with lower-limb loss, provide a unique test of lower-extremity laterality. These groups likely generate a greater limb preference within movement tasks and may have much greater differences in task performance when the task is completed with the non-preferred limb. The discordance between limb preferences and limb dominance, as expressed by the Waterloo Footedness Questionnaire or by self-report, and the high incidence of "mixed-footedness" also question the usefulness of dichotomized definitions of left- or right-limb dominance. Functional evaluations that identify preferences for the task or within balance domains may provide more value to researchers and clinicians when evaluating health, performance, and rehabilitation.

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Appendix A

CHAPTER TWO SUPPLEMENTAL MATERIAL

Table A.1: **Descriptions of participants in chapter two**.

		n	Mean	SD	Max	Min
Age (years)	F – All	7	25.6	4.0	30	20
	F – Included	6	26.3	3.8	30	20
	M – All	7	26.6	5.7	36	20
	M – Included	5	26.7	6.3	36	20
	All	14	26.1	4.8	36	20
	Included	11	27.1	4.7	36	20
Height (cm)		**n**	**Mean**	**SD**	**Max**	**Min**
	F – All	7	170.8	4.4	176.5	165.0
	F – Included	6	169.8	4.0	175.0	165.0
	M – All	7	184.1	8.2	193.0	173.5
	M – Included	5	181.7	8.5	191.5	173.5
	All	14	177.4	9.4	193.0	165.0
	Included	11	175.2	8.7	191.5	165.0
Mass (kg)		**n**	**Mean**	**SD**	**Max**	**Min**
	F – All	7	63.7	9.1	75.5	53.5
	F – Included	6	64.2	9.8	75.5	53.5
	M – All	7	72.4	11.5	90.0	58.0
	M – Included	5	71.5	13.9	90.0	58.0
	All	14	68.0	10.9	90.0	53.5
	Included	11	67.5	11.9	90.0	53.5
BMI ($kg \cdot m^{-2}$)		**n**	**Mean**	**SD**	**Max**	**Min**
	F – All	7	21.8	2.8	25.5	19.0
	F – Included	6	22.2	2.9	25.5	19.0
	M – All	7	21.2	1.9	24.5	19.0
	M – Included	5	21.5	2.2	24.5	19.0
	All	14	21.5	2.3	25.5	19.0
	Included	11	21.9	2.5	25.5	19.0

Note: Participant means, standard deviations, maximums, and minimums for age (years), height (centimeters), mass (kilograms), and body mass index ($kg{\cdot}m^{-2}$). Values shown separately and combined for all female and male participants. Participant values shown for all those recruited and all those included in the analysis.

Table A.2: **Means and standard deviations for margin of stability measures**.

Measure	Perturbation Condition	Walking Speed	Limb	Mean	SD
Posterior Margin of Stability At Foot Strike (% Height)	Anterior (simulated trips)	Fast ($1.0\ stat{\cdot}s^{-1}$)	Left	14.28	1.67
			Right	14.13	1.67
		Estimated Preferred ($0.8\ stat{\cdot}s^{-1}$)	Left	9.22	1.74
			Right	9.08	1.87
		Slow ($0.6\ stat{\cdot}s^{-1}$)	Left	5.29	1.62
			Right	5.14	1.87
	No Perturbations	Fast ($1.0\ stat{\cdot}s^{-1}$)	Left	12.60	1.75
			Right	12.77	1.78
		Estimated Preferred ($0.8\ stat{\cdot}s^{-1}$)	Left	8.40	1.70
			Right	8.37	1.78
		Slow ($0.6\ stat{\cdot}s^{-1}$)	Left	4.13	1.93
			Right	4.08	2.00
	Posterior (simulated slips)	Fast ($1.0\ stat{\cdot}s^{-1}$)	Left	14.28	2.06
			Right	14.44	1.70
		Estimated Preferred ($0.8\ stat{\cdot}s^{-1}$)	Left	9.85	1.88
			Right	9.84	1.94
		Slow ($0.6\ stat{\cdot}s^{-1}$)	Left	6.10	1.87
			Right	6.04	1.68

Table A.2 continued.

Anterior Margin of Stability at Mid-Swing (% Height)	Anterior (simulated trips)	Fast (1.0 stat·s^{-1})	Left	-26.79	1.80
			Right	-27.27	1.36
		Estimated Preferred (0.8 stat·s^{-1})	Left	-19.91	1.01
			Right	-20.16	1.12
		Slow (0.6 stat·s^{-1})	Left	-13.13	1.55
			Right	-13.83	1.17
	No Perturbations	Fast (1.0 stat·s^{-1})	Left	-26.56	1.27
			Right	-26.33	1.25
		Estimated Preferred (0.8 stat·s^{-1})	Left	-19.98	1.11
			Right	-19.84	1.05
		Slow (0.6 stat·s^{-1})	Left	-13.22	1.30
			Right	-13.19	1.00
	Posterior (simulated slips)	Fast (1.0 stat·s^{-1})	Left	-25.95	1.59
			Right	-27.23	1.38
		Estimated Preferred (0.8 stat·s^{-1})	Left	-20.02	1.50
			Right	-20.65	1.38
		Slow (0.6 stat·s^{-1})	Left	-13.95	0.93
			Right	-14.17	1.20

Note: Means and standard deviations for all combinations of perturbation types (simulated trips, none, or simulated slips), walking speed (slow, estimated preferred, or fast), and stepping limb (foot strike) / stance limb (mid-swing).

Table A.3: **All factorial ANOVA results for posterior margin of stability at foot strike.**

MoS	Factorial ANOVA	p	η^2	Post-hoc (I and J)	p	(I - J)	SE
Posterior Margin of Stability At Foot Strike (% Height)	Perturbation Condition	*<0.001	0.766	Anterior and None	*<0.001	1.132	0.209
				Posterior and None	*<0.001	1.697	0.259
				Anterior and Posterior	*0.017	-0.565	0.162
	Walking Speed	*<0.001	0.991	Slow and Normal	*<0.001	-4.000	0.124
				Normal and Fast	*<0.001	-4.623	0.183
				Slow and Fast	*<0.001	-8.623	0.241
	Stepping Limb	0.895	0.002				
	Condition * Speed	0.062	0.197				
	Condition * Limb	0.456	0.075				
	Speed * Limb	0.439	0.079				
	Condition * Speed * Limb	0.778	0.042				

Note: All results from the factorial ANOVA shown above. P-value significance ($p < 0.05$) indicated with '*'. Effect size indicated with partial eta squared ($\eta 2$). Post-hoc comparisons expressed as mean difference and standard error.

Table A.4: **All factorial ANOVA results for anterior margin of stability at mid-swing.**

MoS	Factorial ANOVA	p	η^2	Post-hoc (I and J)	p	(I - J)	SE
Anterior Margin of Stability at Mid-Swing (% Height)	Perturbation Condition	*0.023	0.313				
	Walking Speed	*<0.001	0.996				
	Stepping Limb	*0.023	0.421				
	Speed * Condition	*0.008	0.285	Fast Walking, Anterior and None	0.053	-0.586	0.208
				Fast Walking, Posterior and None	0.923	-0.143	0.247
				Fast Walking, Anterior and Posterior	0.460	-0.443	0.312
				Est. Pref. Walking, Anterior and None	0.750	-0.124	0.132
				Est. Pref. Walking, Posterior and None	0.248	-0.428	0.229
				Est. Pref. Walking, Anterior and Posterior	0.415	0.304	0.203
				Slow Walking, Anterior and None	0.638	-0.276	0.246
				Slow Walking, Posterior and None	*0.019	-0.858	0.250
				Slow Walking, Anterior and Posterior	*0.029	0.582	0.183

Table A.4 continued.

Anterior Margin of Stability at Mid-Swing (% Height)	Condition * Speed	*0.008	0.285	Anterior Perturbations, Slow and Est. Pref.	*<0.001	6.553	0.218
				Anterior Perturbations, Est. Pref. and Fast	*<0.001	6.999	0.306
				Anterior Perturbations, Slow and Fast	*<0.001	13.552	0.262
				No Perturbations, Slow and Est. Pref.	*<0.001	6.705	0.114
				No Perturbations, Est. Pref. and Fast	*<0.001	6.537	0.134
				No Perturbations, Slow and Fast	*<0.001	13.242	0.212
				Posterior Perturbations, Slow and Est. Pref.	*<0.001	6.275	0.159
				Posterior Perturbations, Est. Pref. and Fast	*<0.001	6.252	0.342
				Posterior Perturbations, Slow and Fast	*<0.001	12.527	0.323

Table A.4 continued.

Anterior Margin of Stability at Mid-Swing (% Height)	Limb * Condition	*0.006	0.398	Stance on Non-Dominant Limb, Anterior and None	0.999	-0.027	0.182
				Stance on Non-Dominant Limb, Posterior and None	0.995	-0.056	0.259
				Stance on Non-Dominant Limb, Anterior and Posterior	1.000	0.029	0.288
				Stance on Dominant Limb, Anterior and None	*0.005	-0.631	0.150
				Stance on Dominant Limb, Posterior and None	*0.001	-0.897	0.17
				Stance on Dominant Limb, Anterior and Posterior	*0.032	0.266	0.085
	Condition * Limb			Anterior Perturbations, Dominant and Non-Dominant	*0.027	-0.475	0.183
				No Perturbations, Dominant and Non-Dominant	0.309	0.130	0.121
				Posterior Perturbations, Dominant and Non-Dominant	*0.016	-0.711	0.245
	Speed * Limb	0.371	0.094				
	Condition * Speed * Limb	0.067	0.193				

Note: All results from the factorial ANOVA shown above. P-value significance ($p < 0.05$) indicated with '*'. Effect size indicated with partial eta squared (η^2). Post-hoc comparisons expressed as mean difference and standard error.

Table A.5: **Means and standard deviations for gait parameters.**

Measure	Perturbation Condition	Walking Speed	Limb	Mean	SD
Step Length (cm)	Anterior (simulated trips)	Fast (1.0 stat·s^{-1})	Left	82.5	4.2
			Right	82.9	4.2
		Est. Pref. (0.8 stat·s^{-1})	Left	72.4	3.1
			Right	72.8	3.2
		Slow (0.6 stat·s^{-1})	Left	61.4	2.4
			Right	61.6	2.7
	No Perturbations	Fast (1.0 stat·s^{-1})	Left	84.4	4.1
			Right	84.8	4.4
		Est. Pref. (0.8 stat·s^{-1})	Left	74.1	3.6
			Right	74.1	3.6
		Slow (0.6 stat·s^{-1})	Left	63.1	2.8
			Right	63.4	3.3
	Posterior (simulated slips)	Fast (1.0 stat·s^{-1})	Left	82.1	4.5
			Right	82.6	4.4
		Est. Pref. (0.8 stat·s^{-1})	Left	72.0	3.4
			Right	72.7	4.0
		Slow (0.6 stat·s^{-1})	Left	60.4	2.7
			Right	61.0	3.0
Step Rate (steps·s^{-1})	Anterior (simulated trips)	Fast (1.0 stat·s^{-1})	Left	2.13	0.05
			Right	2.12	0.06
		Est. Pref. (0.8 stat·s^{-1})	Left	1.94	0.04
			Right	1.93	0.04
		Slow (0.6 stat·s^{-1})	Left	1.71	0.06
			Right	1.71	0.04
	No Perturbations	Fast (1.0 stat·s^{-1})	Left	2.08	0.04
			Right	2.07	0.06
		Est. Pref. (0.8 stat·s^{-1})	Left	1.89	0.06
			Right	1.89	0.05
		Slow (0.6 stat·s^{-1})	Left	1.67	0.07
			Right	1.66	0.06
	Posterior (simulated slips)	Fast (1.0 stat·s^{-1})	Left	2.14	0.05
			Right	2.12	0.06
		Est. Pref. (0.8 stat·s^{-1})	Left	1.95	0.04
			Right	1.93	0.05
		Slow (0.6 stat·s^{-1})	Left	1.75	0.06
			Right	1.73	0.07

Table A.5 continued.

Step Width (cm)	Anterior (simulated trips)	Fast (1.0 stat·s^{-1})	Left	11.8	2.1
			Right	11.3	2.2
		Est. Pref. (0.8 stat·s^{-1})	Left	12.0	1.8
			Right	11.8	1.7
		Slow (0.6 stat·s^{-1})	Left	11.7	2.1
			Right	11.9	1.9
	No Perturbations	Fast (1.0 stat·s^{-1})	Left	11.1	2.1
			Right	11.4	2.0
		Est. Pref. (0.8 stat·s^{-1})	Left	11.5	2.0
			Right	11.6	2.1
		Slow (0.6 stat·s^{-1})	Left	11.3	1.7
			Right	11.6	1.7
	Posterior (simulated slips)	Fast (1.0 stat·s^{-1})	Left	11.4	2.0
			Right	11.6	1.9
		Est. Pref. (0.8 stat·s^{-1})	Left	11.6	1.8
			Right	11.7	2.0
		Slow (0.6 stat·s^{-1})	Left	12.0	1.8
			Right	12.0	2.0
Stance Time (% of gait cycle)	Anterior (simulated trips)	Fast (1.0 stat·s^{-1})	Left	63.4	0.5
			Right	63.6	0.4
		Est. Pref. (0.8 stat·s^{-1})	Left	64.9	0.4
			Right	64.8	0.5
		Slow (0.6 stat·s^{-1})	Left	66.7	0.5
			Right	66.7	0.7
	No Perturbations	Fast (1.0 stat·s^{-1})	Left	63.4	0.5
			Right	63.6	0.5
		Est. Pref. (0.8 stat·s^{-1})	Left	64.9	0.5
			Right	65.0	0.6
		Slow (0.6 stat·s^{-1})	Left	66.8	0.7
			Right	66.9	0.7
	Posterior (simulated slips)	Fast (1.0 stat·s^{-1})	Left	63.3	0.5
			Right	63.5	0.4
		Est. Pref. (0.8 stat·s^{-1})	Left	64.8	0.6
			Right	64.8	0.6
		Slow (0.6 stat·s^{-1})	Left	66.3	0.5
			Right	66.4	0.4

Table A.5 continued.

Swing Time (% of gait cycle)	Anterior (simulated trips)	Fast (1.0 stat·s^{-1})	Left	36.6	0.5
			Right	36.4	0.4
		Est. Pref. (0.8 stat·s^{-1})	Left	35.1	0.4
			Right	35.2	0.5
		Slow (0.6 stat·s^{-1})	Left	33.3	0.5
			Right	33.3	0.7
	No Perturbations	Fast (1.0 stat·s^{-1})	Left	36.6	0.5
			Right	36.4	0.5
		Est. Pref. (0.8 stat·s^{-1})	Left	35.1	0.5
			Right	35.0	0.6
		Slow (0.6 stat·s^{-1})	Left	33.2	0.7
			Right	33.1	0.7
	Posterior (simulated slips)	Fast (1.0 stat·s^{-1})	Left	36.7	0.5
			Right	36.5	0.4
		Est. Pref. (0.8 stat·s^{-1})	Left	35.2	0.6
			Right	35.2	0.6
		Slow (0.6 stat·s^{-1})	Left	33.7	0.5
			Right	33.6	0.4
Initial Double Support (% of gait cycle)	Anterior (simulated trips)	Fast (1.0 stat·s^{-1})	Left	13.6	0.4
			Right	13.3	0.4
		Est. Pref. (0.8 stat·s^{-1})	Left	14.9	0.4
			Right	14.7	0.5
		Slow (0.6 stat·s^{-1})	Left	16.8	0.6
			Right	16.5	0.7
	No Perturbations	Fast (1.0 stat·s^{-1})	Left	13.7	0.4
			Right	13.3	0.6
		Est. Pref. (0.8 stat·s^{-1})	Left	15.0	0.4
			Right	14.9	0.8
		Slow (0.6 stat·s^{-1})	Left	17.0	0.5
			Right	16.8	1.0
	Posterior (simulated slips)	Fast (1.0 stat·s^{-1})	Left	13.7	0.3
			Right	13.2	0.4
		Est. Pref. (0.8 stat·s^{-1})	Left	15.0	0.5
			Right	14.4	0.6
		Slow (0.6 stat·s^{-1})	Left	16.6	0.6
			Right	16.3	0.5

Table A.5 continued.

Terminal Double Support (% of gait cycle)	Anterior (simulated trips)	Fast (1.0 stat·s^{-1})	Left	13.3	0.4
			Right	13.6	0.4
		Est. Pref. (0.8 stat·s^{-1})	Left	14.8	0.6
			Right	14.9	0.5
		Slow (0.6 stat·s^{-1})	Left	16.5	0.9
			Right	16.8	0.6
	No Perturbations	Fast (1.0 stat·s^{-1})	Left	13.2	0.6
			Right	13.6	0.4
		Est. Pref. (0.8 stat·s^{-1})	Left	14.8	0.7
			Right	14.9	0.4
		Slow (0.6 stat·s^{-1})	Left	16.7	1.1
			Right	17.0	0.6
	Posterior (simulated slips)	Fast (1.0 stat·s^{-1})	Left	13.2	0.4
			Right	13.7	0.3
		Est. Pref. (0.8 stat·s^{-1})	Left	14.6	0.6
			Right	15.0	0.4
		Slow (0.6 stat·s^{-1})	Left	16.1	0.5
			Right	16.5	0.6
Total Double Support (% of gait cycle)	Anterior (simulated trips)	Fast (1.0 stat·s^{-1})	Left	26.8	0.7
			Right	26.9	0.7
		Est. Pref. (0.8 stat·s^{-1})	Left	29.9	0.8
			Right	29.6	0.7
		Slow (0.6 stat·s^{-1})	Left	33.3	0.9
			Right	33.3	0.9
	No Perturbations	Fast (1.0 stat·s^{-1})	Left	27.0	0.8
			Right	26.9	0.8
		Est. Pref. (0.8 stat·s^{-1})	Left	29.9	1.0
			Right	29.9	1.0
		Slow (0.6 stat·s^{-1})	Left	33.8	1.3
			Right	33.8	1.2
	Posterior (simulated slips)	Fast (1.0 stat·s^{-1})	Left	26.9	0.7
			Right	26.8	0.5
		Est. Pref. (0.8 stat·s^{-1})	Left	29.5	0.9
			Right	29.5	0.8
		Slow (0.6 stat·s^{-1})	Left	32.7	0.7
			Right	32.8	0.7

Note: Means and standard deviations for all combinations of perturbation types (simulated trips, none, or simulated slips), walking speed (slow, estimated preferred, or fast), and stepping limb (left or right).

Table A.6: **All factorial ANOVA results for gait parameters.**

Measure	Factorial ANOVA	p	η^2	Post-hoc (I and J)	p	(I - J)	SE
Step Length (m)	Perturbation Condition	*<0.001	0.676	Anterior and None	*0.005	-0.017	0.004
				Posterior and None	*0.001	-0.022	0.004
				Anterior and Posterior	0.101	0.005	0.002
	Walking Speed	*<0.001	0.992	Slow and Est. Pref.	*<0.001	-0.112	0.004
				Est. Pref. and Fast	*<0.001	-0.102	0.004
				Slow and Fast	*<0.001	-0.214	0.006
	Stepping Limb	0.231	0.140				
	Condition * Speed	0.723	0.049				
	Condition * Limb	0.308	0.111				
	Speed * Limb	0.95	0.005				
	Condition * Speed * Limb	0.543	0.073				
Step Rate (steps·s^{-1})	Perturbation Condition	*<0.001	0.672	Anterior and None	*0.004	0.045	0.010
				Posterior and None	*0.002	0.060	0.012
				Anterior and Posterior	0.103	-0.015	0.006
	Walking Speed	*<0.001	0.992	Slow and Est. Pref.	*<0.001	-0.216	0.008
				Est. Pref. and Fast	*<0.001	-0.186	0.008
				Slow and Fast	*<0.001	-0.402	0.008
	Stepping Limb	0.231	0.140				
	Condition * Speed	0.533	0.063				
	Condition * Limb	0.252	0.129				
	Speed * Limb	0.979	0.002				
	Condition * Speed * Limb	0.619	0.062				

Table A.6 continued.

Step Width (m)	Perturbation Condition	0.105	0.202				
	Walking Speed	0.307	0.108				
	Stepping Limb	0.450	0.058				
	Condition * Speed	0.655	0.058				
	Condition * Limb	0.075	0.229				
	Speed * Limb	0.511	0.065				
	Condition * Speed * Limb	0.335	0.105				

Table A.6 continued.

Stance Time (% of gait cycle)	Perturbation Condition	*0.038	0.279				
	Walking Speed	*<0.001	0.989				
	Stepping Limb	0.526	0.041				
	Speed * Condition	*0.009	0.281	Fast Walking, Anterior and None	0.997	0.012	0.065
				Fast Walking, Posterior and None	0.922	-0.044	0.075
				Fast Walking, Anterior and Posterior	0.787	0.056	0.064
				Est. Pref. Walking, Anterior and None	0.906	-0.089	0.142
				Est. Pref. Walking, Posterior and None	0.643	-0.116	0.104
				Est. Pref. Walking, Anterior and Posterior	0.980	0.027	0.074
				Slow Walking, Anterior and None	0.386	-0.154	0.099
				Slow Walking, Posterior and None	0.053	-0.491	0.174
				Slow Walking, Anterior and Posterior	*0.049	0.337	0.117

Table A.6 continued.

Stance Time (% of gait cycle)	Condition * Speed	*0.009	0.281	Anterior Perturbations, Slow and Est. Pref.	*<0.001	1.857	0.114
				Anterior Perturbations, Est. Pref. and Fast	*<0.001	1.377	0.109
				Anterior Perturbations, Slow and Fast	*<0.001	3.234	0.131
				No Perturbations, Slow and Est. Pref.	*<0.001	1.922	0.070
				No Perturbations, Est. Pref. and Fast	*<0.001	1.478	0.067
				No Perturbations, Slow Walking and Fast	*<0.001	3.400	0.116
				Posterior Perturbations, Slow and Est. Pref.	*<0.001	1.547	0.135
				Posterior Perturbations, Est. Pref. and Fast	*<0.001	1.406	0.063
				Posterior Perturbations, Slow and Fast	*<0.001	2.953	0.112
	Condition * Limb	0.772	0.026				
	Speed * Limb	0.240	0.133				
	Condition * Speed * Limb	0.866	0.031				

Table A.6 continued.

Swing Time (% of gait cycle)	Perturbation Condition	*0.038	0.279				
	Walking Speed	*<0.001	0.989				
	Stepping Limb	0.526	0.041				
	Speed * Condition	*0.009	0.281	Fast Walking, Anterior and None	0.997	-0.012	0.065
				Fast Walking, Posterior and None	0.922	0.044	0.075
				Fast Walking, Anterior and Posterior	0.787	-0.056	0.064
				Est. Pref. Walking, Anterior and None	0.906	0.089	0.142
				Est. Pref. Walking, Posterior and None	0.643	0.116	0.104
				Est. Pref. Walking, Anterior and Posterior	0.980	-0.027	0.074
				Slow Walking, Anterior and None	0.386	0.154	0.099
				Slow Walking, Posterior and None	0.053	0.491	0.174
				Slow Walking, Anterior and Posterior	*0.049	-0.337	0.117

Table A.6 continued.

Swing Time (% of gait cycle)	Condition * Speed	*0.009	0.281	Anterior Perturbations, Slow and Est. Pref.	*<0.001	-1.857	0.114
				Anterior Perturbations, Est. Pref. and Fast	*<0.001	-1.377	0.109
				Anterior Perturbations, Slow and Fast	*<0.001	-3.234	0.131
				No Perturbations, Slow and Est. Pref.	*<0.001	-1.922	0.070
				No Perturbations, Est. Pref. and Fast	*<0.001	-1.478	0.067
				No Perturbations, Slow Walking and Fast	*<0.001	-3.400	0.116
				Posterior Perturbations, Slow and Est. Pref.	*<0.001	-1.547	0.135
				Posterior Perturbations, Est. Pref. and Fast	*<0.001	-1.406	0.063
				Posterior Perturbations, Slow and Fast	*<0.001	-2.953	0.112
	Condition * Limb	0.772	0.026				
	Speed * Limb	0.240	0.133				
	Condition * Speed * Limb	0.866	0.031				

Table A.6 continued.

Initial Double Support (% of gait cycle)	Perturbation Condition	*0.025	0.308				
	Walking Speed	*<0.001	0.990				
	Stepping Limb	0.061	0.307				
	Speed * Condition	*0.019	0.251	Fast Walking, Anterior and None	0.999	0.010	0.090
				Fast Walking, Posterior and None	0.908	-0.045	0.073
				Fast Walking, Anterior and Posterior	0.692	0.055	0.053
				Est. Pref. Walking, Anterior and None	0.671	-0.143	0.134
				Est. Pref. Walking, Posterior and None	0.195	-0.228	0.112
				Est. Pref. Walking, Anterior and Posterior	0.363	0.085	0.075
				Slow Walking, Anterior and None	0.093	-0.209	0.084
				Slow Walking, Posterior and None	0.058	-0.453	0.164
				Slow Walking, Anterior and Posterior	0.140	0.244	0.109

Table A.6 continued.

Initial Double Support (% of gait cycle)	Condition * Speed	*0.019	0.251	Anterior Perturbations, Slow and Est. Pref.	*<0.001	1.853	0.107
				Anterior Perturbations, Est. Pref. and Fast	*<0.001	1.345	0.084
				Anterior Perturbations, Slow and Fast	*<0.001	3.198	0.115
				No Perturbations, Slow and Est. Pref.	*<0.001	1.919	0.054
				No Perturbations, Est. Pref. and Fast	*<0.001	1.498	0.072
				No Perturbations, Slow Walking and Fast	*<0.001	3.417	0.104
				Posterior Perturbations, Slow and Est. Pref.	*<0.001	1.694	0.127
				Posterior Perturbations, Est. Pref. and Fast	*<0.001	1.316	0.064
				Posterior Perturbations, Slow and Fast	*<0.001	3.010	0.113
	Condition * Limb	0.383	0.085				
	Speed * Limb	0.589	0.052				
	Condition * Speed * Limb	0.321	0.108				

Table A.6 continued.

Terminal Double Support (% of gait cycle)	Perturbation Condition	0.145	0.175				
	Walking Speed	*<0.001	0.988				
	Stepping Limb	0.063	0.305				
	Speed * Condition	*0.001	0.355	Fast Walking, Anterior and None	0.999	-0.011	0.085
				Fast Walking, Posterior and None	1.000	0.003	0.089
				Fast Walking, Anterior and Posterior	0.995	-0.014	0.060
				Est. Pref. Walking, Anterior and None	1.000	0.004	0.130
				Est. Pref. Walking, Posterior and None	0.951	-0.046	0.094
				Est. Pref. Walking, Anterior and Posterior	0.895	0.050	0.077
				Slow Walking, Anterior and None	0.393	-0.179	0.116
				Slow Walking, Posterior and None	0.085	-0.508	0.200
				Slow Walking, Anterior and Posterior	0.065	0.329	0.122

Table A.6 continued.

Terminal Double Support (% of gait cycle)	Condition * Speed	*0.001	0.355	Anterior Perturbations, Slow and Est. Pref.	*<0.001	1.803	0.110
				Anterior Perturbations, Est. Pref. and Fast	*<0.001	1.442	0.108
				Anterior Perturbations, Slow and Fast	*<0.001	3.246	0.120
				No Perturbations, Slow and Est. Pref.	*<0.001	1.987	0.103
				No Perturbations, Est. Pref. and Fast	*<0.001	1.427	0.067
				No Perturbations, Slow Walking and Fast	*<0.001	3.414	0.124
				Posterior Perturbations, Slow and Est. Pref.	*<0.001	1.525	0.127
				Posterior Perturbations, Est. Pref. and Fast	*<0.001	1.378	0.064
				Posterior Perturbations, Slow and Fast	*<0.001	2.903	0.108
	Condition * Limb	0.644	0.043				
	Speed * Limb	0.376	0.085				
	Condition * Speed * Limb	0.967	0.014				

Table A.6 continued.

Total Double Support (% of gait cycle)	Perturbation Condition	*0.017	0.335				
	Walking Speed	*<0.001	0.992				
	Stepping Limb	0.774	0.009				
	Speed * Condition	*0.004	0.311	Fast Walking, Anterior and None	0.828	-0.109	0.137
				Fast Walking, Posterior and None	0.832	-0.091	0.115
				Fast Walking, Anterior and Posterior	0.996	-0.018	0.091
				Est. Pref. Walking, Anterior and None	0.961	-0.116	0.257
				Est. Pref. Walking, Posterior and None	0.212	-0.375	0.19
				Est. Pref. Walking, Anterior and Posterior	0.428	0.259	0.175
				Slow Walking, Anterior and None	0.089	-0.487	0.194
				Slow Walking, Posterior and None	*0.042	-0.994	0.335
				Slow Walking, Anterior and Posterior	0.065	0.507	0.188

Table A.6 continued.

Total Double Support (% of gait cycle)	Condition * Speed	*0.004	0.311	Anterior Perturbations, Slow and Est. Pref.	*<0.001	3.541	0.185
				Anterior Perturbations, Est. Pref. and Fast	*<0.001	2.917	0.182
				Anterior Perturbations, Slow and Fast	*<0.001	6.458	0.185
				No Perturbations, Slow and Est. Pref.	*<0.001	3.912	0.130
				No Perturbations, Est. Pref. and Fast	*<0.001	2.924	0.126
				No Perturbations, Slow Walking and Fast	*<0.001	6.836	0.212
				Posterior Perturbations, Slow and Est. Pref.	*<0.001	3.293	0.247
				Posterior Perturbations, Est. Pref. and Fast	*<0.001	2.640	0.134
				Posterior Perturbations, Slow and Fast	*<0.001	5.933	0.193
	Condition * Limb	0.881	0.013				
	Speed * Limb	0.449	0.077				
	Condition * Speed * Limb	0.231	0.128				

Note: Note: All results from the factorial ANOVA shown above. P-value significance ($p < 0.05$) indicated with '*'. Effect size indicated with partial eta squared (η^2). Post-hoc comparisons expressed as mean difference and standard error.

Waterloo Footedness Questionnaire—Revised [128]

Participants answer the following questions with the foot (left, right, or equal) that they would use to perform each activity, and, if applicable, the extent to which they would use that foot (always or usually).

1. Which foot would you use to kick a stationary ball at a target straight in front of you?
2. If you had to stand on one foot, which foot would it be?
3. Which foot would you use to smooth sand at the beach?
4. If you had to step up onto a chair, which foot would you place on the chair first?
5. Which foot would you use to stomp on a fast-moving bug?
6. If you were to balance on one foot on a railway track, which foot would you use?
7. If you wanted to pick up a marble with your toes, which foot would you use?
8. If you had to hop on one foot, which foot would you use?
9. Which foot would you use to help push a shovel into the ground?
10. During relaxed standing, people initially put most of their weight on one foot, leaving the other leg slightly bent. Which foot do you put most of your weight on first?
11. Is there any reason (i.e. injury) why you have changed your foot preference for any of the above activities?
12. Have you ever been given special training or encouragement to use a particular foot for certain activities?
13. If you have answered **YES** for either question 11 or 12, please explain:

Figure A.1: **Waterloo Footedness Questionnaire—Revised.** Participants respond to ten questions, five related to "stability" tasks and five related to "mobility" tasks, with which limb they would perform a task (left, right, or equal) and the extent to which they would use that limb for the task (always or usually). Questions from Elias and colleagues [128].

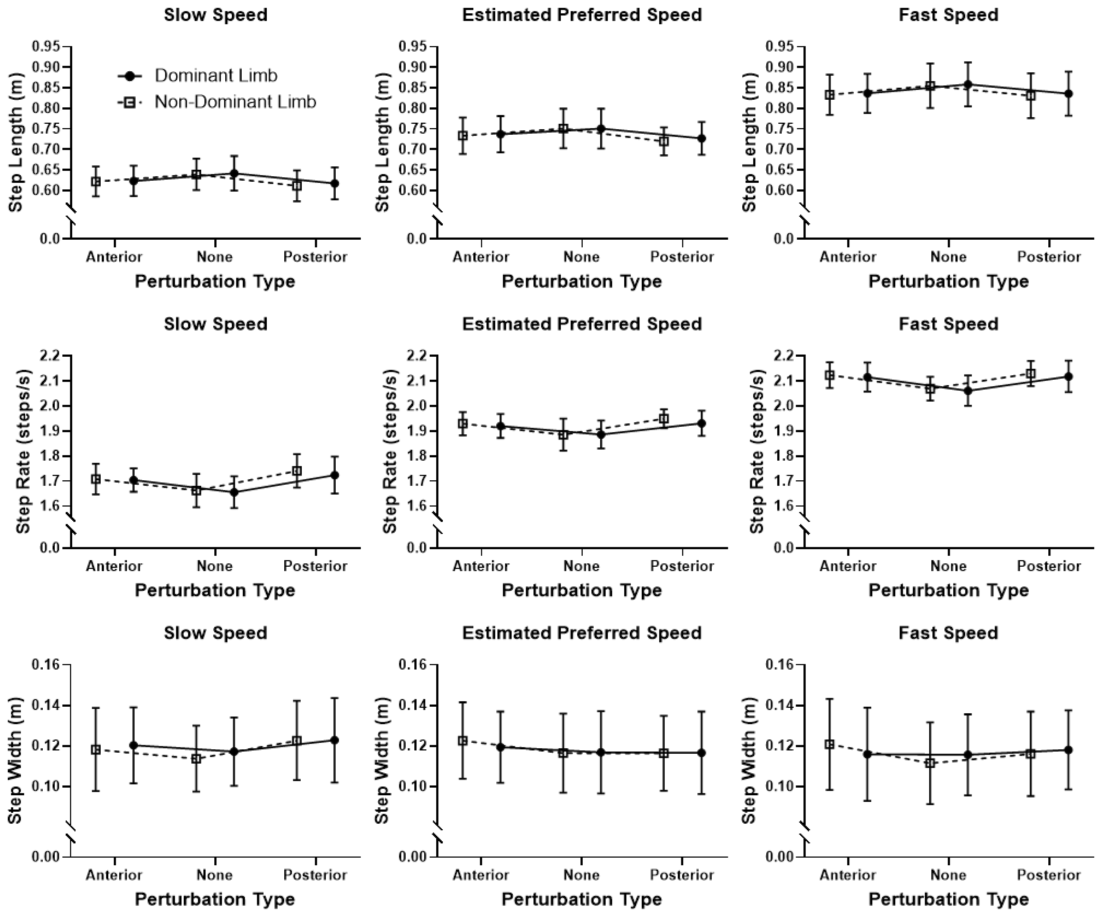

Figure A.2: **Summary results for step length, step rate, and step width.** The dominant limb shown with solid circles and solid lines. The non-dominant limb shown with open squares and dashed lines. Gait parameter means and standard deviations shown across perturbation types (simulated trips, none, or simulated slips) and walking speeds (slow, estimated preferred, or fast). <u>Top Row</u> – Participants decreased step length for trials with perturbations compared to trials without perturbations and increased step length with increasing walking speed. <u>Middle Row</u> – Participants increased step rate for trials with perturbations compared to trials without perturbations and increased step rate with increasing walking speed. <u>Bottom Row</u> – Participants did not change step width when threatened with perturbations or when changing walking speeds.

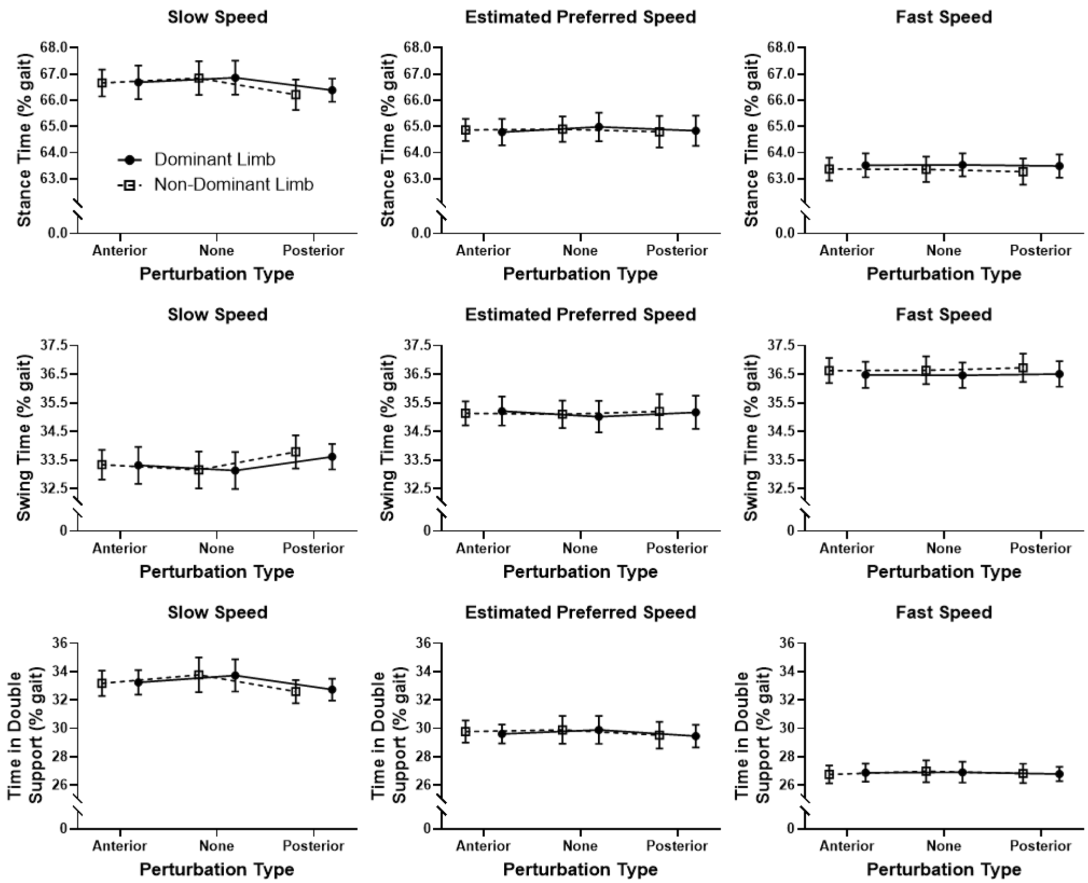

Figure A.3: **Summary results for stance time, swing time, and time in double support.** The dominant limb shown with solid circles and solid lines. The non-dominant limb shown with open squares and dashed lines. Gait parameter means and standard deviations shown across perturbation types (simulated trips, none, or simulated slips) and walking speeds (slow, estimated preferred, or fast). Top Row – With each increase in walking speed, participants decreased their percentage of time spent in stance. Middle Row – With each increase in walking speed, participants increased their percentage of time spent in swing. Bottom Row – A decrease in the percent of time spent in double support was observed when threatened with posterior perturbations at the slow walking speed. With each increase in walking speed, participants decreased their percentage of time spent in double support.

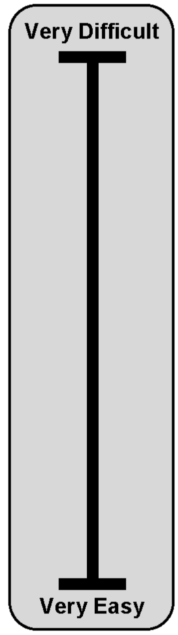

Figure A.4: **Visual analog scale for perceived trial difficulty.** Participants responded on a visual analog scale to the prompt “Make a mark indicating how difficult you perceived the trial you just completed” where the responses could range from Very Easy (scored 0) to Very Difficult (scored 100). Specifications and scoring for the scale were done according to Cline and colleagues [136].

During this trial, it became [________] to recover from the perturbations over time.

Much Easier	Easier	No Different	Harder	Much Harder

Figure A.5: **Likert scale for perceived change in perturbation recovery difficulty over time.** Participants responded on a five-point Likert scale to the prompt “During this trial, it became ______ to recover from the perturbations over time” where the responses could range from Much Easier (scored -2) to Much Harder (scored +2).

Appendix B

CHAPTER THREE SUPPLEMENTAL MATERIAL

Table B.1: **Descriptions of participants in chapter three**.

		n	Mean	SD	Max	Min
Age (years)	**Female**	15	24.7	3.7	30	19
	Male	15	26.9	6.4	40	18
	All	30	25.8	5.2	40	18
Height (cm)		**n**	**Mean**	**SD**	**Max**	**Min**
	Female	15	169.8	6.7	178.5	155.5
	Male	15	183.4	7.6	196.0	173.0
	All	30	176.6	9.9	196.0	155.5
Mass (kg)		**n**	**Mean**	**SD**	**Max**	**Min**
	Female	15	63.0	9.6	81.0	47.0
	Male	15	75.7	12.2	97.5	60.0
	All	30	69.4	12.6	97.5	47.0
BMI ($kg \cdot m^{-2}$)		**n**	**Mean**	**SD**	**Max**	**Min**
	Female	15	21.9	3.3	31.2	18.4
	Male	15	22.5	3.2	29.1	18.5
	All	30	22.2	3.2	31.2	18.4
Foot Length (cm)		**n**	**Mean**	**SD**	**Max**	**Min**
	Female	15	27.8	1.4	29.5	23.5
	Male	15	30.8	1.4	34.0	29.0
	All	30	29.3	2.1	34.0	23.5

Table B.1 continued.

		n	Mean	SD	Max	Min
Foot Width (cm)	Female	15	10.2	0.6	11.0	8.5
	Male	15	11.0	0.5	11.5	10.0
	All	30	10.6	0.7	11.5	8.5
Left Leg Length (cm)		n	Mean	SD	Max	Min
	Female	15	88.1	5.1	97.0	75.5
	Male	15	94.9	5.5	104.0	86.5
	All	30	91.5	6.3	104.0	75.5
Right Leg Length (cm)		n	Mean	SD	Max	Min
	Female	15	88.1	5.0	96.5	75.5
	Male	15	95.0	5.5	104.0	86.5
	All	30	91.5	6.3	104.0	75.5

Note: Participant means, standard deviations, maximums, and minimums for age (years), height (centimeters), mass (kilograms), body mass index ($kg \cdot m^{-2}$), foot length (cm), foot width (cm), left leg length (cm), and right leg length (cm). Values shown separately and combined for all female and male participants.

Table B.2: **Participant responses to athletic/activity history questionnaire.**

Rank	Sport/Activity	Number of Responses	Percentage of Participants
1	Soccer	20	66.7 %
2	Running (Distance)	18	60.0 %
3	Swimming	16	53.3 %
T4	Baseball/Softball	14	46.7 %
T4	Walking/Hiking	14	46.7 %
T4	Weight Lifting	14	46.7 %
7	Bicycling	13	43.3 %
8	Basketball	12	40.0 %
9	Skiing	10	33.3 %
T10	Driving (Manual Transmission)	8	26.7 %
T10	Gymnastics	8	26.7 %
T10	Track (Field Events)	8	26.7 %
T10	Yoga	8	26.7 %
T14	Dance	7	23.3 %
T14	Football	7	23.3 %
T14	Running (Sprints)	7	23.3 %
T17	Circuit Training	6	20.0 %
T17	Rock Climbing	6	20.0 %
T17	Volleyball	6	20.0 %
T20	Tennis	5	16.7 %
T20	Wrestling	5	16.7 %
T22	Field Hockey	4	13.3 %
T22	Lacrosse	4	13.3 %
T22	Martial Arts	4	13.3 %
T22	Racquetball	4	13.3 %
T26	Archery	3	10.0 %
T26	Bowling	3	10.0 %
T26	Diving	3	10.0 %
T26	Fishing	3	10.0 %
T26	Golf	3	10.0 %
T26	Painting	3	10.0 %
T26	Skateboarding	3	10.0 %
T26	Surfing	3	10.0 %
T34	Billiards	2	6.7 %
T34	Table Tennis	2	6.7 %
T34	Rowing/Crew	2	6.7 %
T34	Rugby	2	6.7 %

Note: The athletic/activity history questionnaire was compiled from the Compendium of Physical Activities – Updated [152] and the CARE Consortium Baseline Questionnaire [153]. A minimum of two responses was used to make this list.

Table B.3: **Means, standard deviations, minimums, and maximums for inter-limb asymmetry values for each task.**

Measure	**Mean ILA**	**SD**	**Min**	**Max**
Minimum Lateral MoS (%BH)	-0.28	0.79	-2.45	1.42
Postural Sway - Eyes Opened Beta (%)	-9.93	22.60	-61.1	26.5
Postural Sway - Eyes Closed Beta (%)	-4.63	22.21	-50.2	34.7
Gait Initiation - AP COP Displacement APA Phase (%BH)	-0.06	0.62	-1.29	1.10
Gait Initiation - ML COP Displacement APA Phase (%BH)	0.21	0.70	-1.40	1.51
Simulated Trips - COM to Step Length (%BH)	0.34	3.11	-5.64	5.66
Simulated Trips - COM to Step Width (%BH)	-0.69	2.16	-4.14	4.88
Simulated Slips - COM to Step Length (%BH)	0.59	5.53	-9.99	11.01
Simulated Slips - COM to Step Width (%BH)	0.06	2.57	-5.96	6.00

Note: Inter-limb asymmetry (ILA) calculated according to equation 3.1. A negative ILA indicates a greater value on the right limb, while a positive ILA indicates a greater value on the left limb.

Table B.4: **Left and right limb task measurement means, standard deviations, minimums, and maximums.**

Measure	Mean	SD	Min	Max
Left Minimum Lateral MoS (%BH)	1.05	0.71	-0.28	2.31
Right Minimum Lateral MoS (%BH)	1.33	0.70	-0.47	2.77
Postural Sway - Eyes Opened Left Beta (β)	0.90	0.11	0.70	1.12
Postural Sway - Eyes Opened Right Beta (β)	1.00	0.14	0.68	1.34
Postural Sway - Eyes Closed Left Beta (β)	0.93	0.12	0.63	1.19
Postural Sway - Eyes Closed Right Beta (β)	0.97	0.11	0.77	1.23
Gait Initiation - Left AP COP Displacement APA Phase (cm)	2.91	1.95	0.46	8.21
Gait Initiation - Right AP COP Displacement APA Phase (cm)	3.02	2.07	0.45	10.05
Gait Initiation - Left ML COP Displacement APA Phase (cm)	2.76	1.13	0.28	5.00
Gait Initiation - Right ML COP Displacement APA Phase (cm)	2.39	1.09	0.28	4.55
Simulated Trips - Left Step COM to Step Length (%BH)	13.9	3.4	6.8	21.5
Simulated Trips - Right Step COM to Step Length (%BH)	13.7	4.0	1.2	19.1
Simulated Trips - Left Step COM to Step Width (%BH)	15.2	3.6	5.7	20.6
Simulated Trips - Right Step COM to Step Width (%BH)	5.5	1.6	2.3	8.1
Simulated Slips - Left Step COM to Step Length (%BH)	15.7	5.5	6.4	25.2
Simulated Slips - Right Step COM to Step Length (%BH)	15.2	3.6	5.7	20.6
Simulated Slips - Left Step COM to Step Width (%BH)	3.2	2.8	-0.4	8.7
Simulated Slips - Right Step COM to Step Width (%BH)	3.4	2.2	-1.7	7.5

Table B.5: **Preferred and non-preferred stepping limb task measurement means, standard deviations, minimums, and maximums.**

Measure	Mean	SD	Min	Max
Gait Initiation - Preferred AP COP Displacement APA Phase (cm)	3.18	1.68	0.67	7.81
Gait Initiation - Non-Preferred AP COP Displacement APA Phase (cm)	3.19	2.34	0.49	10.05
Gait Initiation - Preferred ML COP Displacement APA Phase (cm)	2.41	1.00	0.71	5.00
Gait Initiation - Non-Preferred ML COP Displacement APA Phase (cm)	2.88	1.06	1.16	4.56
Simulated Trips - Preferred Step COM to Step Length (%BH)	13.2	3.0	6.8	19.1
Simulated Trips - Non-Preferred Step COM to Step Length (%BH)	14.5	4.3	1.2	21.5
Simulated Trips - Preferred Step COM to Step Width (%BH)	5.4	1.6	1.5	8.5
Simulated Trips - Non-Preferred Step COM to Step Width (%BH)	5.0	1.8	1.3	8.2
Simulated Slips - Preferred Step COM to Step Length (%BH)	14.3	4.0	6.4	20.8
Simulated Slips - Non-Preferred Step COM to Step Length (%BH)	17.0	5.0	5.7	25.2
Simulated Slips - Preferred Step COM to Step Width (%BH)	3.9	2.2	-0.4	7.5
Simulated Slips - Non-Preferred Step COM to Step Width (%BH)	2.5	2.7	-1.7	8.7

Note: Stepping tasks were completed with the self-selected limb for each trial. Preferred and non-preferred stepping limbs were determined by the majority choice for six trials, and no limb preference was determined if both limbs were self-selected equally. Four participants had no preference during gait initiation. All participants had a preference during simulated trips. One participant had no preference during simulated slips.

Appendix C

INTER-LIMB ASYMMETRY EXPLANATION

The Inter-limb Asymmetry (ILA) measures the between-limb difference and scales that value to the participant.

$$ILA = \left(\frac{Left\ Limb - Right\ Limb}{scaling\ factor}\right) \cdot 100 \qquad \text{Equation 3.1}$$

A negative percentage indicates greater value on the right limb, while a positive percentage indicates greater value on the left limb. The ILA was a modification of the asymmetry indices described in Carpes et al. [114] and is expressed in units of percent of scaling factor. We chose to modify the asymmetry indices for three reasons:

1. The MoS can be either a positive or negative value; therefore, some indices can be undefined with a denominator of zero if the limb values are equal in size but opposite in sign – see $SI_{\%}$ equation 1 and $ASI_{\%}$ equation 2 of Carpes et al. [114] recreated in Equation C.1 and Equation C.2 – whereas the ILA cannot be undefined.

$$SI_{\%} = \left[\frac{(Right - Left)}{\frac{(Right + Left)}{2}}\right] \cdot 100 \qquad \text{Equation C.1}$$

$$ASI_{\%} = \left[\frac{|X_r - X_l|}{\frac{1}{2}(X_r + X_l)}\right] \cdot 100 \qquad \text{Equation C.2}$$

2. Both previous indices are influenced by the magnitude of each input value instead of the magnitude of the difference between values (i.e. if both values are large, then the percentage value will be smaller than if there were the same inter-limb difference but with both values small). If participant 1 $X_{r1} = 1$ and $X_{l1} = 2$,

participant 2 X_{r2} = 10 and X_{l2} = 11, and the scaling factor = 170.2 cm (height), then

$$SI_{\%1} = \left[\frac{(1-2)}{\frac{1+2}{2}}\right] \cdot 100 = -66.6\% \quad \text{Equation C.3}$$

$$SI_{\%2} = \left[\frac{(10-11)}{\frac{10+11}{2}}\right] \cdot 100 = -9.5\% \quad \text{Equation C.4}$$

$$ASI_{\%1} = \left[\frac{|1-2|}{\frac{1}{2}(1+2)}\right] \cdot 100 = 66.6\% \quad \text{Equation C.5}$$

$$ASI_{\%2} = \left[\frac{|10-11|}{\frac{1}{2}(10+11)}\right] \cdot 100 = 9.5\% \quad \text{Equation C.6}$$

Whereas, the ILA maintains consistency with the magnitude of difference.

$$ILA_1 = \left(\frac{(2-1)}{170.2}\right) \cdot 100 = 0.59\% \text{ BH} \quad \text{Equation C.7}$$

$$ILA_2 = \left(\frac{(11-10)}{170.2}\right) \cdot 100 = 0.59\% \text{ BH} \quad \text{Equation C.8}$$

3. Both indices will report the same asymmetry if the ratio between the two values is the same (i.e. two participants would have the same percentage value if they both have an inter-limb ratio of two but different inter-limb differences). If participant 1 X_{r1} = 2 and X_{l1} = 4, participant 2 X_{r2} = 10 and X_{l2} = 20, and the scaling factor = 170.2 cm (height), then

$$SI_{\%1} = \left[\frac{(2-4)}{\frac{2+4}{2}}\right] \cdot 100 = -66.6\% \quad \text{Equation C.9}$$

$$SI_{\%2} = \left[\frac{(10-20)}{\frac{10+20}{2}}\right] \cdot 100 = -66.6\% \quad \text{Equation C.10}$$

$$ASI_{\%1} = \left[\frac{|2-4|}{\frac{1}{2}(2+4)}\right] \cdot 100 = 66.6\% \quad \text{Equation C.11}$$

$$ASI_{\%2} = \left[\frac{|10-20|}{\frac{1}{2}(10+20)}\right] \cdot 100 = 66.6\% \qquad \text{Equation C.12}$$

Whereas, the ILA shows differences even when the ratio between limbs is the same between participants.

$$ILA_1 = \left(\frac{(4-2)}{170.2}\right) \cdot 100 = 1.18\%\ BH \qquad \text{Equation C.13}$$

$$ILA_2 = \left(\frac{(20-10)}{170.2}\right) \cdot 100 = 5.88\%\ BH \qquad \text{Equation C.14}$$

For these three reasons, the modified ILA value will be used to represent the inter-limb asymmetry more effectively across a wide range of contexts.

Appendix D

INSTITUTIONAL REVIEW BOARD APPROVALS – CHAPTER TWO

UNIVERSITY OF DELAWARE

Institutional Review Board
210H Hullihen Hall
Newark, DE 19716
Phone: 302-831-2137
Fax: 302-831-2828

DATE:	May 13, 2019
TO:	Jeremy Crenshaw, PhD
FROM:	University of Delaware IRB
STUDY TITLE:	[1436920-1] Anteroposterior walking stability with threats to balance
SUBMISSION TYPE:	New Project
ACTION:	APPROVED
EFFECTIVE DATE:	May 13, 2019
NEXT REPORT DUE:	May 12, 2020
REVIEW TYPE:	Expedited Review
REVIEW CATEGORY:	Expedited review category # *(4,6,7)*

Thank you for your New Project submission to the University of Delaware Institutional Review Board (UD IRB). The UD IRB has reviewed and APPROVED the proposed research and submitted documents via Expedited Review in compliance with the pertinent federal regulations.

As the Principal Investigator for this study, you are responsible for, and agree that:

- All research must be conducted in accordance with the protocol and all other study forms as approved in this submission. Any revisions to the approved study procedures or documents must be reviewed and approved by the IRB prior to their implementation. Please use the UD amendment form to request the review of any changes to approved study procedures or documents.
- Informed consent is a process that must allow prospective participants sufficient opportunity to discuss and consider whether to participate. IRB-approved and stamped consent documents must be used when enrolling participants and a written copy shall be given to the person signing the informed consent form.
- Unanticipated problems, serious adverse events involving risk to participants, and all non-compliance issues must be reported to this office in a timely fashion according with the UD requirements for reportable events. All sponsor reporting requirements must also be followed.

The UD IRB REQUIRES the submission of a PROGRESS REPORT DUE ON May 12, 2020. A continuing review/progress report form must be submitted to the UD IRB at least 45 days prior to the due date to allow for the review of that report.

If you have any questions, please contact the UD IRB Office at (302) 831-2137 or via email at hsrb-research@udel.edu. Please include the study title and reference number in all correspondence with this office.

Institutional Review Board
210H Hullihen Hall
Newark, DE 19716
Phone: 302-831-2137
Fax: 302-831-2828

DATE:	January 27, 2020
TO:	Jeremy Crenshaw, PhD
FROM:	University of Delaware IRB
STUDY TITLE:	[1436920-2] Anteroposterior walking stability with threats to balance
SUBMISSION TYPE:	Amendment/Modification
ACTION:	APPROVED
EFFECTIVE DATE:	January 27, 2020
NEXT REPORT DUE:	May 12, 2020
REVIEW TYPE:	Expedited Review
REVIEW CATEGORY:	Expedited review category # *(4,6,7)*

Thank you for your Amendment/Modification submission to the University of Delaware Institutional Review Board (UD IRB). The UD IRB has reviewed and APPROVED the proposed research and submitted documents via Expedited Review in compliance with the pertinent federal regulations.

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Institutional Review Board
210H Hullihen Hall
Newark, DE 19716
Phone: 302-831-2137
Fax: 302-831-2828

DATE:	May 11, 2020
TO:	Jeremy Crenshaw, PhD
FROM:	University of Delaware IRB
STUDY TITLE:	[1436920-3] Anteroposterior walking stability with threats to balance
SUBMISSION TYPE:	Continuing Review/Progress Report
ACTION:	APPROVED
EFFECTIVE DATE:	May 11, 2020
NEXT REPORT DUE:	May 12, 2021
REVIEW TYPE:	Administrative Review
REVIEW CATEGORY:	Expedited review category # *(2,6,7)*

Thank you for your Continuing Review/Progress Report submission to the University of Delaware Institutional Review Board (UD IRB). The UD IRB has reviewed and APPROVED the proposed research and submitted documents via Administrative Review in compliance with the pertinent federal regulations.

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The UD IRB REQUIRES the submission of a PROGRESS REPORT DUE ON May 12, 2021. A continuing review/progress report form must be submitted to the UD IRB at least 45 days prior to the due date to allow for the review of that report.

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Institutional Review Board
210H Hullihen Hall
Newark, DE 19716
Phone: 302-831-2137
Fax: 302-831-2828

DATE:	May 11, 2021
TO:	Jeremy Crenshaw, PhD
FROM:	University of Delaware IRB
STUDY TITLE:	[1436920-4] Anteroposterior walking stability with threats to balance
SUBMISSION TYPE:	Continuing Review/Progress Report
ACTION:	APPROVED
EFFECTIVE DATE:	May 11, 2021
NEXT REPORT DUE:	May 12, 2022
REVIEW TYPE:	Administrative Review
REVIEW CATEGORY:	Expedited review category # *(4,6,7)*

Thank you for your Continuing Review/Progress Report submission to the University of Delaware Institutional Review Board (UD IRB). The UD IRB has reviewed and APPROVED the proposed research and submitted documents via Administrative Review in compliance with the pertinent federal regulations.

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The UD IRB REQUIRES the submission of a PROGRESS REPORT DUE ON May 12, 2022. A continuing review/progress report form must be submitted to the UD IRB at least 45 days prior to the due date to allow for the review of that report.

If you have any questions, please contact the UD IRB Office at (302) 831-2137 or via email at hsrb-research@udel.edu. Please include the study title and reference number in all correspondence with this office.

INSTITUTIONAL REVIEW BOARD

www.udel.edu

Institutional Review Board
210H Hullihen Hall
Newark, DE 19716
Phone: 302-831-2137
Fax: 302-831-2828

DATE:	June 22, 2021
TO:	Jeremy Crenshaw, PhD
FROM:	University of Delaware IRB
STUDY TITLE:	[1436920-5] Anteroposterior walking stability with threats to balance
SUBMISSION TYPE:	Amendment/Modification
ACTION:	APPROVED
EFFECTIVE DATE:	June 22, 2021
NEXT REPORT DUE:	May 12, 2022
REVIEW TYPE:	Expedited Review
REVIEW CATEGORY:	Expedited review category # *(4,6,7)*

Thank you for your Amendment/Modification submission to the University of Delaware Institutional Review Board (UD IRB). The UD IRB has reviewed and APPROVED the proposed research and submitted documents via Expedited Review in compliance with the pertinent federal regulations.

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The UD IRB REQUIRES the submission of a PROGRESS REPORT DUE ON May 12, 2022. A continuing review/progress report form must be submitted to the UD IRB at least 45 days prior to the due date to allow for the review of that report.

If you have any questions, please contact the UD IRB Office at (302) 831-2137 or via email at hsrb-research@udel.edu. Please include the study title and reference number in all correspondence with this office.

Appendix E

INSTITUTIONAL REVIEW BOARD APPROVALS – CHAPTER THREE

UNIVERSITY OF DELAWARE

Institutional Review Board
210H Hullihen Hall
Newark, DE 19716
Phone: 302-831-2137
Fax: 302-831-2828

DATE:	August 27, 2019
TO:	Jeremy Crenshaw, PhD
FROM:	University of Delaware IRB
STUDY TITLE:	[1468601-1] A study of balance and mobility strategies across different tasks
SUBMISSION TYPE:	New Project
ACTION:	APPROVED
EFFECTIVE DATE:	August 27, 2019
NEXT REPORT DUE:	August 26, 2020
REVIEW TYPE:	Expedited Review
REVIEW CATEGORY:	Expedited review category # *(4)*

Thank you for your New Project submission to the University of Delaware Institutional Review Board (UD IRB). The UD IRB has reviewed and APPROVED the proposed research and submitted documents via Expedited Review in compliance with the pertinent federal regulations.

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The UD IRB REQUIRES the submission of a PROGRESS REPORT DUE ON August 26, 2020. A continuing review/progress report form must be submitted to the UD IRB at least 45 days prior to the due date to allow for the review of that report.

If you have any questions, please contact the UD IRB Office at (302) 831-2137 or via email at hsrb-research@udel.edu. Please include the study title and reference number in all correspondence with this office.

Institutional Review Board
210H Hullihen Hall
Newark, DE 19716
Phone: 302-831-2137
Fax: 302-831-2828

DATE:	January 27, 2020
TO:	Jeremy Crenshaw, PhD
FROM:	University of Delaware IRB
STUDY TITLE:	[1468601-2] A study of balance and mobility strategies across different tasks
SUBMISSION TYPE:	Amendment/Modification
ACTION:	APPROVED
EFFECTIVE DATE:	January 27, 2020
NEXT REPORT DUE:	August 26, 2020
REVIEW TYPE:	Expedited Review
REVIEW CATEGORY:	Expedited review category # *(4)*

Thank you for your Amendment/Modification submission to the University of Delaware Institutional Review Board (UD IRB). The UD IRB has reviewed and APPROVED the proposed research and submitted documents via Expedited Review in compliance with the pertinent federal regulations.

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If you have any questions, please contact the UD IRB Office at (302) 831-2137 or via email at hsrb-research@udel.edu. Please include the study title and reference number in all correspondence with this office.

Institutional Review Board
210H Hullihen Hall
Newark, DE 19716
Phone: 302-831-2137
Fax: 302-831-2828

DATE:	August 17, 2020
TO:	Jeremy Crenshaw, PhD
FROM:	University of Delaware IRB
STUDY TITLE:	[1468601-3] A study of balance and mobility strategies across different tasks
SUBMISSION TYPE:	Continuing Review/Progress Report
ACTION:	APPROVED
EFFECTIVE DATE:	August 17, 2020
NEXT REPORT DUE:	August 26, 2021
REVIEW TYPE:	Administrative Review
REVIEW CATEGORY:	Expedited review category # *(4)*

Thank you for your Continuing Review/Progress Report submission to the University of Delaware Institutional Review Board (UD IRB). The UD IRB has reviewed and APPROVED the proposed research and submitted documents via Administrative Review in compliance with the pertinent federal regulations.

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- Unanticipated problems, serious adverse events involving risk to participants, and all non-compliance issues must be reported to this office in a timely fashion according with the UD requirements for reportable events. All sponsor reporting requirements must also be followed.
- In-person research interaction with subjects cannot begin until the UD moratorium in response to thedeclaration of national emergency related tothe COVID-19 pandemic is lifted. Please continue to reference https://research.udel.edu/coronavirus for the most up-to-date recommendations.

The UD IRB REQUIRES the submission of a PROGRESS REPORT DUE ON August 26, 2021. A continuing review/progress report form must be submitted to the UD IRB at least 45 days prior to the due date to allow for the review of that report.

If you have any questions, please contact the UD IRB Office at (302) 831-2137 or via email at hsrb-research@udel.edu. Please include the study title and reference number in all correspondence with this office.

Institutional Review Board
210H Hullihen Hall
Newark, DE 19716
Phone: 302-831-2137
Fax: 302-831-2828

DATE:	August 20, 2021
TO:	Jeremy Crenshaw, PhD
FROM:	University of Delaware IRB
STUDY TITLE:	[1468601-4] A study of balance and mobility strategies across different tasks
SUBMISSION TYPE:	Continuing Review/Progress Report
ACTION:	APPROVED
EFFECTIVE DATE:	August 20, 2021
NEXT REPORT DUE:	August 26, 2022
REVIEW TYPE:	Administrative Review
REVIEW CATEGORY:	Expedited review category # *(4)*

Thank you for your Continuing Review/Progress Report submission to the University of Delaware Institutional Review Board (UD IRB). The UD IRB has reviewed and APPROVED the proposed research and submitted documents via Administrative Review in compliance with the pertinent federal regulations.

As the Principal Investigator for this study, you are responsible for, and agree that:

- All research must be conducted in accordance with the protocol and all other study forms as approved in this submission. Any revisions to the approved study procedures or documents must be reviewed and approved by the IRB prior to their implementation. Please use the UD amendment form to request the review of any changes to approved study procedures or documents.
- Informed consent is a process that must allow prospective participants sufficient opportunity to discuss and consider whether to participate. IRB-approved and stamped consent documents must be used when enrolling participants and a written copy shall be given to the person signing the informed consent form.
- Unanticipated problems, serious adverse events involving risk to participants, and all non-compliance issues must be reported to this office in a timely fashion according with the UD requirements for reportable events. All sponsor reporting requirements must also be followed.

The UD IRB REQUIRES the submission of a PROGRESS REPORT DUE ON August 26, 2022. A continuing review/progress report form must be submitted to the UD IRB at least 45 days prior to the due date to allow for the review of that report.

If you have any questions, please contact the UD IRB Office at (302) 831-2137 or via email at hsrb-research@udel.edu. Please include the study title and reference number in all correspondence with this office.