NORTHWESTERN UNIVERSITY

A Stability-Based Approach to Post-Stroke Gait Training

A DISSERTATION

SUBMITTED TO THE GRADUATE SCHOOL

IN PARTIAL FULFILLMENT OF THE REQUIREMENTS

for the degree

DOCTOR OF PHILOSOPHY

Field of Biomedical Engineering

By

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EVANSTON, ILLINOIS

September 2021

Abstract

Humans have a remarkable ability to create stable walking patterns that can resist and recover from perturbations. Unfortunately, this ability is substantially impaired after a stroke, limiting mobility and contributing to a high fall rate. To facilitate gait training during post-stroke rehabilitation, clinicians often incorporate body-weight support (BWS) systems that apply vertical forces to the pelvis and trunk to assist with upright walking. However, these BWS forces change the patient's walking dynamics and balance control, potentially limiting the patient's ability to relearn walking balance. My thesis proposes a new stability-based framework for post-stroke gait rehabilitation, which considers how rehabilitation devices interact with the control of gait stability. To do this, I first explore how a BWS system affects gait stability and stepping coordination in healthy adults. I demonstrate that walking with high levels of BWS reduces active modulation of foot placement to control walking balance and encourages a compensatory stepping strategy that poorly translates to the dynamics of real-world walking. Next, I explore how a new type of rehabilitation device, which applies lateral assistive forces to the body, affects the control of gait stability in adults with and without stroke. I find that individuals with stroke use a stepping strategy that relies on wide steps that are less coordinated to the body state during walking. In addition, I find that walking with lateral assistance does not interfere with the control of stability and tends to improve stepping coordination for individuals with stroke, warranting further investigation. Together, these results; 1) advance our understanding of how common assistive tools may interfere with learning walking balance and, 2) present an alternative method of gait training for adults with stroke. Further, this thesis provides a framework and a path forward for modifying current rehabilitation techniques to better target gait stability.

Acknowledgements

To my advisers:

Drs. Keith Gordon and Jules Dewald for your mentorship and guidance over the past seven years. Keith, for teaching me how to be comfortable with uncertainty in my work and how to appreciate long periods of contemplative silence during meetings. Jules, for pushing me to purse bigger and harder scientific problems and questions.

To my committee members:

Drs. Eric Perreault and Ana Maria Acosta Eric for your valuable feedback on my work. Eric, for teaching me the importance of properly framing and describing my work. Ana Maria, for convincing me to pursue this dual-degree and for your support throughout the program.

To my family, coworkers, friends I have made along the way:

I could not have finished this work without your support.

To my Northwestern friends and classmates, for helping me stay afloat through the many hours of coding and dissecting.

To Kirsten, Roberto, Justin, Carolina, Randy, Marjorie, and all of NUPTHMS, for your encouragement and support, as well as the entertainment during the CSM parties.

To Tara, Natalia, Steve, Mary, Amanda, Geoff, Alex, and all of the Human Agility Lab, your support and helping me become a better scientist.

To Peter, Patricia, Brian, for everything that you have taught me. Brian, for being the best older brother and helping me become a proper gamer and adult. Peter, for teaching me how to better understand the world and my place in it. Patricia, for teaching me how to be curious and empathetic, and how to be colorful with my language.

To Janeen and &, you both mean the world to me. &, you are my most loyal non-human companion. Janeen, for bringing my life happiness, fulfillment, and adventure.

To my funders:

The National Institute of Health (T32HD057845 and T32EB009406), Department of Veterans Affairs (IK2 RX000717 and I01RX001979), the Northwestern Feinberg School of Medicine, and the American taxpayers for providing me with the resources and support to develop into a scientist.

Thank you to everyone.

Preface

Chapter 2: This work has been published in the Journal of Biomechanics [1].

Chapter 3: This work is under review.

Chapter 4: This work has been published in IEEE Transactions on Neural Systems and Rehabilitation Engineering [2].

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Chapter 1: Introduction

Purpose

The purpose of my dissertation is to better understand how devices used for gait training interact with the control of stability. This work is intended to first, understand how external assistance provided through body weight support and lateral forces affect control of gait stability and second, create a framework for a stability-based gait training intervention for people post-stroke.

It is a challenge for individuals post-stroke to create stable walking patterns that can resist and/or recover from perturbations [3], [4]. This impairment limits mobility and contributes to high fall rates, which have significant socioeconomic costs [5]–[7]. Thus, improving walking stability is a priority during post-stroke gait rehabilitation. To allow patients post-stroke to safely practice taking steps, clinicians will often incorporate body-weight support (BWS) devices that apply supporting vertical forces to the body. However, applying forces to the body may change the dynamics of walking and consequently how the patient practices controlling stability. The non-intuitive interaction between BWS devices and the stability control may explain why outcomes from BWS interventions have shown limited benefits when compared with conventional therapy [8], [9].

My first two studies examine how BWS, one of the most common gait rehabilitation tools, affects whole-body center of mass (COM) dynamics and the control strategies used to maintain walking stability. Results of these studies suggest that the use of BWS substantially changes how people control walking stability. My third study examines the effects of a novel rehabilitation device, that applies supporting lateral forces which dampens the velocity of the COM, on the control of walking stability in individuals with stroke. I find that when people post-stroke walk with lateral assistance from a damping field, they tend to improve their stepping coordination, which suggests this type of field can be used as a stability-based gait training method.

These studies advance our understanding of how a common rehabilitation tool may interfere with learning dynamic balance and then suggests an alternative method of gait training. Outcomes from these studies provide both a framework and a path forward for modifying current rehabilitation techniques to better target gait stability.

Motivation and Significance

Human walking is an inherently unstable task. To remain upright during walking, we must balance our precariously-positioned upper body above a pelvis that rhythmically oscillates in multiple planes as it vaults, like an inverted pendulum (Figure 1.1), over a semi-rigid support limb [10]. Our bipedal gait pattern, which alternates between periods of single- and double-limb support, creates regular episodes of instability when the COM travels beyond the base of support (BOS), only to be alleviated as we step and establish a new BOS [11]–[13]. While bipedal gait can appear to be an entirely automatic process, it requires active coordination by the locomotor system to effectively reposition the limbs each step to support, propel, and control motion of the COM [14].

“Stable” gait can withstand perturbations while achieving a target trajectory [12], [15]. The challenge with controlling human walking is that every step creates a collision with the ground, generating peak forces exceeding bodyweight that accelerate the COM and create substantial rotational moments about the joints of the supporting limb [11], [16]. A combination of neuromotor noise and changes in the external environment result in variable collision dynamics each step. If the COM deviates too far from an ideal trajectory in the frontal plane, gravitational moments can act to increasingly accelerate the COM away from the BOS [12], [17]. To maintain stability, humans can adjust the state of the COM (i.e., position and velocity) through muscular moments created by the stance limb or modulate their future BOS by adjusting their swing limb trajectory and resulting foot placement location [11], [12].

It can be difficult to measure the control of stability. Gait variability is a common measure, as high variability is often believed to be indicative of a loss of stability [18], [19]. However, it is difficult to interpret how gait variability is controlled by the locomotor system [19] or determine its relationship with fall risk [18], [20]. New methods of linear and nonlinear models have been developed that measure how the locomotor system recovers from small perturbations [15], [17], [19] and large perturbations [15] with the intent to understand how gait stability is controlled in humans. Strong evidence suggests that a primary mechanism for maintaining gait stability is the modulation of lateral foot placement [11], [14], [21], [22].

The use of an appropriate foot placement is critical to maintaining frontal plane gait stability. This is not a straightforward process as there are an infinite number of potential foot placement positions. It has been suggested that foot placement is the result of both passive and active processes [11], [17]. Passive dynamic walking robots can recover from small perturbations without any active coordination due to the intrinsic properties and dynamics of the system [23].

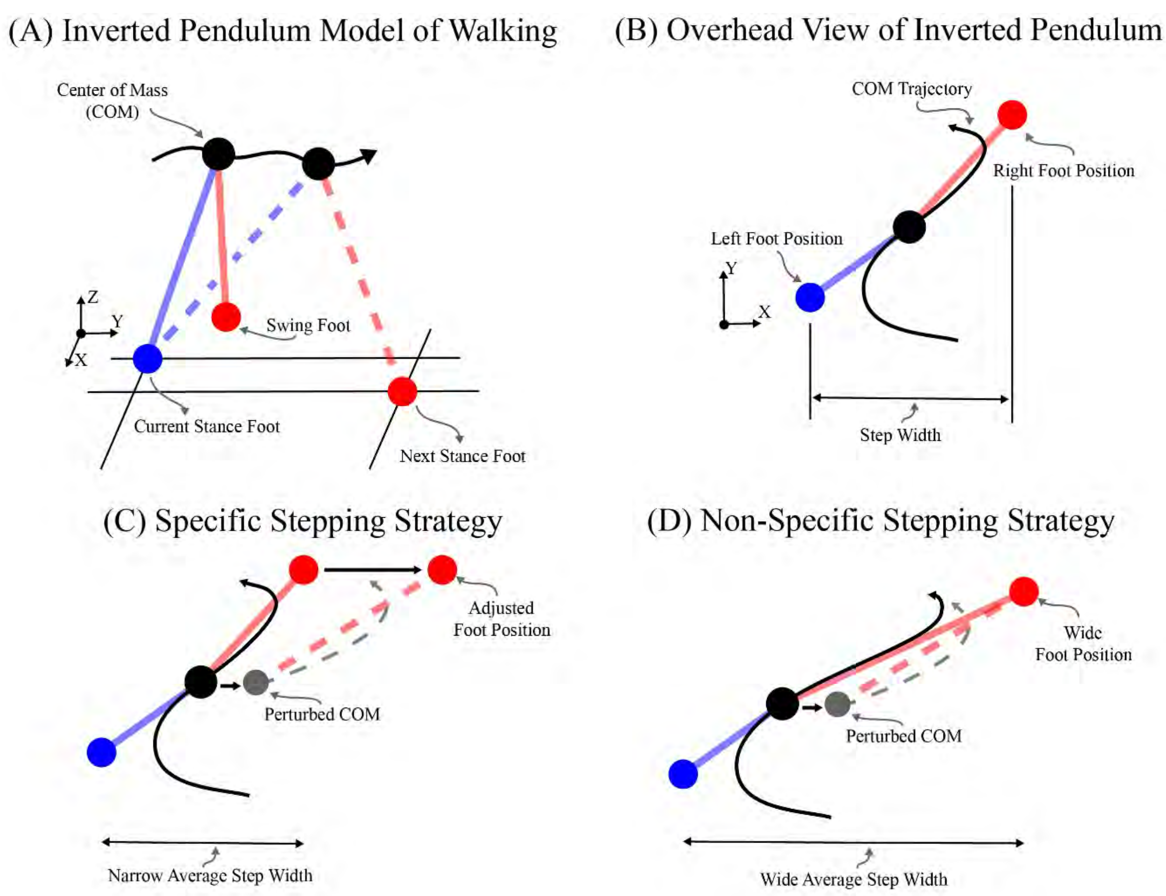

Figure 1.1: Inverted Pendulum Model and Stepping Strategies (A) Diagram of the inverted pendulum model of walking. The coordinate reference frame is located at the center of the stance foot, and the x-, y-, and z-axes are the mediolateral, anteroposterior, and vertical directions, respectively. The solid lines represent the current limb positions, and the dotted lines represent future limb positions. To maintain stability from step-to-step, a new stance foot position must be chosen that prevents the COM from exceeding the BOS. (B) Overhead view of the inverted pendulum model, where step width is defined as the mediolateral distance between the feet. (C) Representation of a specific stepping strategy where there is coordination between the COM and swing limb to create an appropriate foot placement position for the current COM state. As the COM is perturbed to the right, the foot is placed further to the right in response. (D)

Representation of a non-specific stepping strategy where there is a consistently wide step width, which does not adjust in response to a perturbation of the COM.

While humans can similarly exploit the passive dynamics of bipedalism, the control of foot placement requires more active involvement by the human locomotor control system [14]. It is likely that higher and lower-order neural mechanisms (corticospinal pathways and spinal reflexes, respectively) work in conjunction to coordinate foot placement. Experiments that disrupt sensory feedback with visual [14] or vestibular [24] perturbations have demonstrated that foot placement becomes more variable when sensory information is unreliable, which is suggestive of active control by the locomotor system. Recent work has begun to explore how the locomotor system controls foot placement by investigating the stepping strategies that humans use to maintain step-to-step stability.

There are two basic strategies that humans use to coordinate foot placement. The first is a *specific* stepping strategy, where foot placement position is tightly regulated based on the COM state (Figure 1.1) [25], [26]. For instance, on steps where the COM is moving faster and further away from the stance limb, individuals place their swing limb further away from the body. Recent work has found that COM state during swing phase is highly predictive of subsequent foot placement position in healthy young adults [17], and researchers have suggested that this is reflective of active control by the human locomotor system [12], [17], [22]. The second stepping strategy is *non-specific*, where there is reduced coordination between COM state and the swing foot (Figure 1.1) [25], [26]. This is associated with gait patterns that use indiscriminately wide steps, which is common for older adults [27], [28]. There are trade-offs within each of the two strategies [29]. If stepping is actively controlled by the locomotor system, then specific stepping strategies require a reliable estimate of the COM state and precise control of swing limb trajectory. Non-specific strategies are more energetically costly [30]–[32] and appear to be the preferred strategy for determining foot placement following injuries to the central nervous system [33], [34]. Young adults switch between strategies when exposed to perturbations [35] or less demanding conditions [36]. These experiments suggest that stepping strategies are modifiable and potentially learnable, which are enticing findings for neurorehabilitation.

Following a neurologic injury, such as a hemiparetic stroke, the abilities to precisely control foot placement [37], [38] or sense the body state [39] are substantially impaired. A stroke interrupts descending motor commands and ascending sensory information, limiting the ability of the locomotor system to actively coordinate stepping. As a result, individuals with stroke tend to use non-specific stepping strategies with wide step widths [40] and greater variability [18] than has been observed among healthy adults. Stroke survivors at a high-risk for falls have reduced coordination compared to low-risk stroke survivors [3], [41], and the paretic extremity is less coordinated with the COM than the non-paretic extremity [33]. In addition, individuals with stroke have difficulty targeting their foot placements [38] and matching prescribed step widths [37] during gait. These findings suggest that there is a post-stroke shift towards non-specific stepping strategies. However, recent work has suggested that stepping strategies remain modifiable for individuals with chronic stroke [42], and it may be possible to design interventions to specifically target stepping coordination for individuals with stroke and ultimately improve gait stability.

BWS systems are one of the most common tools used for post-stroke gait training [43]–[45]. These systems allow patients with weakness and instability to practice walking in a controlled and safe environment [45]. The main advantage of BWS treadmill training is that it can progressively challenge a patient’s abilities, by modulating either gait speed or the muscular effort required to stay upright. As a result, patients with severe post-stroke locomotor deficits can initiate gait training within hours after the injury. These systems can be used overground if patients are able to initiate steps independently; otherwise, they are most used in conjunction with a treadmill. This environment allows for repeated practice of stepping but may discourage active control of stability. BWS systems work by applying a vertical force through the COM to offset the effects of gravity. They are typically fixed overhead [8], [46] or move in one degree of freedom [47], and as a result, they may generate lateral forces that change walking dynamics [46], [48]. Thus, while the assistance provided by BWS has benefits to rehabilitation, it may also encourage less coordinated stepping or even compensatory behaviors that are detrimental to real-world balance control.

Unfortunately, current rehabilitation interventions may interfere with learning or training new stepping strategies for individuals with stroke. Many post-stroke gait interventions focus on improving gait speed [43], endurance [43], or symmetry [49], rather than stepping coordination. Rehabilitation devices, such as BWS, have been designed to provide progressive gait and balance training for individuals with stroke [50]; however, they have had mixed effects [9]. BWS treadmill training is no more effective than conventional therapy at improving clinical measures of gait and balance [9], [43]. There are several issues with these devices, which are namely that they do not reproduce the body dynamics in normal walking [48], [51], can constrain movement

[52], [53], and decrease the need to volitionally control stepping [54]. It is reasonable to suspect that BWS changes the demands to maintain stability, which may affect stepping coordination and limit post-stroke functional improvement. The lack of improvement in walking balance following BWS interventions is concerning for the field of gait rehabilitation and introduces the need for a new approach to improve post-stroke gait stability.

Currently, it is unclear how BWS interacts with gait stability. Several researchers have suggested that the overhead suspension of BWS provides a ‘stabilizing’ effect in the frontal plane [46], [55], [56]. Experiments have found that adults adopt narrower steps [46] with less trunk variability [55], suggesting that the demands to maintain stability are lower with BWS. However, BWS has also been found to increase kinematic [52] and kinetic [57] variability, suggestive of an increased demand to maintain stability. As mentioned, it is difficult to interpret the relationship between gait variability and stability. To my knowledge, the interaction between BWS and gait stability has yet to be measured in any population. This leaves researchers and clinicians without an understanding of how the most common post-stroke gait training intervention affects a basic requirement of gait. For my first aim, I will investigate the relationship between BWS and gait stability.

In addition, it is unclear how BWS interacts with stepping coordination. There is reason to suspect that BWS will lead to changes in stepping behaviors used to control frontal plane balance. The vertical forces of BWS are applied nonuniformly to the body [46], [56], [58], leading to differences in inertial properties between the trunk and legs. BWS alters the step length [51], [58], step timing [51], [58], inter-joint coordination[51], kinematic trajectories [53], and kinetic trajectories [57], which likely affect how the body coordinates control of foot placement. To my knowledge, the interaction between BWS, foot placement, and stepping coordination has yet to be investigated in any population. The lack of research in this area has serious clinical implications, as the stepping strategies that are learned during gait training with BWS may poorly relate to real-world walking or worse may even translate unstable behaviors. For my second aim, I will investigate how BWS affects stepping coordination.

Moving forward, a new stability-based approach is needed for post-stroke gait training. New interventions that aim to modify stepping coordination should be specifically targeted to challenge the control of stability. Using the principles of activity-dependent neuroplasticity [59], [60], interventions should allow for repeated practice of stepping strategies and provide adaptable task intensity. This approach will provide guidance to clinicians on how best to use gait training devices, as well as a framework for developing new stability-based gait training interventions.

Using a stability-based framework for post-stroke gait rehabilitation, rehabilitation interventions should target, modulate, and sufficiently challenge the control of gait stability. Researchers have developed a new type of gait training device that applies lateral forces to the COM during treadmill walking as a form of gait training [34], [61]–[64]. It is theorized that these devices can manipulate the demands to actively control stability by providing progressive lateral assistance. My laboratory has developed a cable-driven robotic device, the Agility Trainer [61], that applies lateral forces to the COM as individuals walk on a treadmill. My colleagues have demonstrated that the Agility Trainer can produce ‘stabilizing’ and ‘destabilizing’ force fields [62], [63], [65], which lead to changes in stepping strategy for individuals with incomplete spinal cord injury [63], [65]. The stabilizing field, or Damping Field, applies a force that opposes the real-time lateral velocity of the COM, and may encourage practice of more regulated stepping.

The lateral assistance provided by these devices is an exciting direction for post-stroke gait rehabilitation, which warrants further investigation. There is reason to suspect that individuals with stroke will also adapt their control of stability when walking in a Damping Field created by the Agility Trainer. Previous research has found that external lateral stabilization [36], which constrains lateral COM position, reduces post-stroke stepping coordination [66]. Recent work has demonstrated that a laterally-dampened, multi-directional BWS system can improve locomotor control in individuals with stroke [56], [67]. It is unclear how individuals with or without stroke adapt their stepping coordination in a Damping Field. While post-stroke stepping coordination has been compared between the paretic and non-paretic extremities [33], it is necessary to compare these individuals to those without stroke in order to examine any higher-order or between-limb deficits in stepping coordination. For my third aim, I expand upon our current understanding of post-stroke gait coordination and the effects of lateral assistance on post-stroke gait.

Specific Aims and Summary of Experiments

Specific Aim 1: To investigate how body-weight support systems affect gait stability.

Hypothesis: BWS will change the demands to maintain stability, which will scale with the level of BWS.

Healthy young adults walked on a treadmill with increasing levels of BWS. I examined how BWS level affected general (step width mean and variability) and dynamic (margin of stability [13], [68]) measures of gait stability. I found a complex interaction between BWS and gait stability.

As the level of BWS increased, participants decreased their step width variability, which is traditionally indicative of a more stable gait pattern. However, participants also increased their step width, which is associated with a less stable gait pattern. There were no changes in margin of stability.

Walking with high levels of BWS induces complex changes in walking dynamics that do not replicate the active control requirements needed for real-world community ambulation.

Specific Aim 2: To investigate how body-weight support systems affect stepping coordination.

Hypothesis: BWS will reduce coordination between the COM state and foot placement, which will scale with the level of BWS. Walking with BWS will encourage a non-specific stepping strategy.

Healthy young adults walked on a treadmill with varying levels of BWS at their preferred gait speed. I investigated how BWS affects walking dynamics by measuring the effects of BWS on COM state, foot placement, and stepping coordination. Stepping coordination was quantified using the correlation between lateral COM state during the swing-phase of gait and the subsequent foot placement location.

I found that as the level of BWS increased, participants adopted a less coordinated stepping strategy and exhibited asymmetries in COM motion and foot placement position. This work suggests that BWS encourages a compensatory, non-specific stepping strategy, which may minimize practice of active control of gait stability.

Specific Aim 3: To determine how individuals with stroke adapt their stepping coordination when walking in a damping force field.

Hypothesis: Walking with a damping force field will change the demands to maintain stability and reduce stepping coordination for individuals with and without stroke. Individuals with stroke will have reduced coordination compared to individuals without stroke.

Adults with and without hemiparetic stroke walked on a treadmill in normal conditions, as well as in a Damping Field that applied lateral forces to the pelvis that were proportional in magnitude and opposite in direction to the real-time lateral COM velocity. I investigated how the Damping Field affected general measures of gait stability (step width mean and variability) and stepping coordination.

I found that individuals with stroke had a less coordinated stepping strategy and wider step widths than individuals without stroke, suggesting that individuals with stroke use a more non-specific stepping strategy. The Damping Field did not affect stepping coordination for individuals without stroke but tended to increase the correlation between COM state and foot placement for individuals with stroke.

Individuals with stroke used a stepping strategy that did not strictly regulate foot placement relative to the COM state but rather relied on wide steps for gait stability. This work suggests that lateral assistive forces can be used to create a safe environment for gait training and may improve stepping coordination in individuals with stroke. Future work should develop interventions that encourage specific stepping strategies as a form of stability-based gait training.

Chapter 2: Body Weight Support Impacts Lateral Stability during Treadmill Walking

Abstract

Body weight support (BWS) systems are a common tool used in gait rehabilitation. BWS systems may alter the requirements for an individual to actively stabilize by 1) providing lateral restoring forces that reduce the requirements for the nervous system to actively stabilize and 2) altering the natural passive dynamics of walking by reducing the effect of gravity, which could increase the requirements to actively stabilize. The goal of the current study was to quantify the interaction between BWS and lateral stability. We hypothesized that when able-bodied people walk with BWS: 1) the lateral restoring forces provided by BWS would reduce the requirements to stabilize in the frontal plane when comparing dynamically similar gaits, and 2) increasing BWS would decrease the stabilizing gravitational moment in the frontal plane and increase the requirements to stabilize when speed is constrained. Our findings partly support these hypotheses but indicate a complex interaction between BWS and lateral stability. With BWS, subjects significantly decreased step width variability and significantly increased step width ($p < 0.05$) for both the dynamically similar and speed-matched conditions. The decrease in step width variability may be attributable to a combination of lateral restoring forces decreasing the mechanical requirements to stabilize and an enhanced sense of position that could have improved locomotor control. Increases in step width when walking with high levels of BWS could have been due to changes in passive dynamics, which increased the demands on the nervous system for controlling walking stability in multiple planes.

Introduction

The restoration of locomotor balance after neurological injury is critical for successful independent walking [69], [70]. A body weight support (BWS) system is a common tool used to facilitate locomotor training after spinal cord injury [71] and stroke [72] by reducing the muscle force required to counteract gravity. However, BWS exerts forces on an individual that could change the requirements of the nervous system to actively stabilize the body during gait. For BWS to be utilized effectively, it is important to understand how its implementation changes the requirements of locomotor stability. Here we focus specifically on lateral stability because simple models suggest that human walking is passively unstable in the frontal plane and requires active control [23], [73].

BWS could change the requirements to control frontal plane locomotor stability through two mechanisms (Figure 2.1). First, BWS systems that use cable-driven actuation routed through an overhead suspension could provide lateral restoring forces [74] that decrease the requirements to actively control lateral stability. If the suspension restricts lateral translations, the BWS system will act as a pendulum and pull the user beneath the fulcrum [46]. These lateral restoring forces may act as external stabilizers, reducing the requirements of the user to actively stabilize in the frontal plane. The lateral restoring forces provided by BWS has been estimated to be as high as 8.5 N when walking with 50% BWS [75]. Previous studies showing that people take narrower steps [46] and increase frontal plane trunk acceleration regularity [48], [76] with BWS support the idea that BWS provides external lateral stabilization.

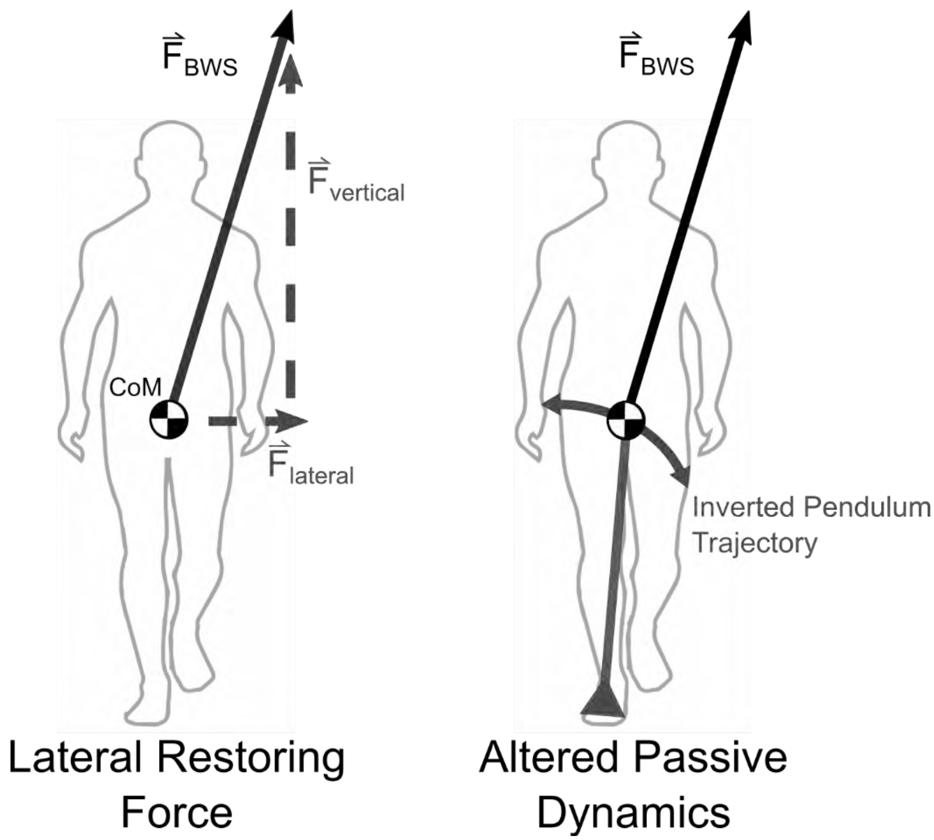

Figure 2.1: Effect of BWS on Frontal Plane Dynamics (Left) Schematic of the lateral restoring force, $F_{lateral}$, provided by the BWS support system as the subject deviates laterally. We expect this force to stabilize the subject and reduce the requirements to actively stabilize in the frontal plane. (Right) Schematic of the altered frontal plane balance control when walking with BWS. Without BWS, the subject controls balance at the subtalar joint through a combination of gravitational moments, joint center accelerations, and inversion/eversion muscle moments [11]; with BWS, the gravitational moment is reduced due to the vertical force of the BWS and an additional lateral moment through the COM is introduced. We expect these adaptations to frontal plane balance to require active control, and increase the requirements to stabilize in the frontal plane.

Gravity itself creates a stabilizing moment about the stance limb(s) that aids in maintaining the center of mass (CoM) within the base of support during walking [11]. Reducing this stabilizing gravitational moment with BWS may increase the requirements for active lateral balance control. During single support, the whole body lateral inertial moment about the subtalar joint is counteracted directly by a combination of a gravitational moment, subtalar joint center accelerations, and inversion/eversion muscle moments [11]. The gravitational moment is greater than 40 Nm during single support and is controlled primarily by changes in the distance of the stance foot from the COM [11]. Thus, when gravity is reduced with BWS, the requirements to actively control lateral inertial moments during walking should increase. Previous human research has shown that sagittal plane gait kinematic [52] and kinetic [57] variability increase with BWS, suggesting that reducing gravity may indeed have a destabilizing effect.

To investigate these two competing effects of BWS independently, we propose that the potentially destabilizing effect of reducing the gravitational moment about the support limb can be minimized by observing gait with varying levels of BWS during dynamically similar walking conditions. The principle of dynamic similarity states that geometrically similar bodies utilizing pendulum mechanics will have similar gait dynamics when the ratio between gravitational and inertial forces are equivalent [10], [77]. The value of this ratio is known as the Froude number (Fr). To maintain a constant Fr, the inertia forces must be reduced proportionally which can be achieved by decreasing walking speed [77]. While Fr has been used primarily to examine control of sagittal plane pendulum dynamics, there is evidence that lateral accelerations during walking will also decrease as walking speed is reduced [78]. The stabilizing effect of the gravitational moment which counteracts lateral inertial forces should be similar when walking in dynamically similar conditions as the ratio of gravitational to inertial forces will be constant.

We observed the effects of BWS on frontal plane stability during walking in two conditions, Dynamically Similar and Speed-Matched. In the first condition, walking speed was adjusted to produce dynamically similar (constant Fr) walking patterns across BWS levels. In this condition, stability changes should be driven by the lateral restoring force provided by BWS as the ratio of gravitational to inertial forces will be constant across conditions. We hypothesized that in Dynamically Similar conditions increasing BWS would decrease the requirements to actively stabilize in the frontal plane due to the relative increase in lateral restoring forces provided by the BWS. In a second condition, walking speed was kept constant across all BWS levels. In this Speed-Matched condition, we hypothesized an increase in the requirements for active lateral stabilization because the stabilizing effect of gravity to control the lateral CoM inertia would be decreased as BWS increased and that this effect would be greater than the stabilizing effects of BWS.

Methods

Subjects

Eight healthy subjects participated in the experiment. Subjects gave written informed consent, and the experimental protocol was approved by the Northwestern University Institutional Review Board. Subjects were 26 ± 4 years, height 1.74 ± 0.12 m, mass 71.4 ± 10.2 kg, leg length 0.88 ± 0.08 m, and 3 males / 5 females.

Experimental setup

All walking occurred on an oversized treadmill (1.39 m belt width) with no handrails (Tuff Tread, Willis, TX). Subjects wore a trunk harness attached to a motorized overhead BWS system (Aretech, Ashburn, VA). The BWS system's actuated trolley maintained a fore-aft position directly above the subject but did not track lateral position.

12 retro-reflective markers were placed bilaterally on the pelvis and feet. Markers were affixed to the 2nd and 5th metatarsals, calcaneus, greater trochanters, and anterior and posterior superior iliac spines. A 10-camera motion capture system (Qualisys, Gothenurg, Sweden) recorded 3D marker positions at 100 Hz.

Protocol

Subjects performed 7 walking trials at different levels of BWS and speed, each of which included a 5-minute accommodation period that was followed immediately by a 2-minute data collection period. The order of the trials was randomized, and subjects were given rest as needed between trials. BWS and speed were selected to compare the effect of changing BWS during Dynamically Similar and Speed-Matched conditions.

Dynamically Similar Conditions

During the Dynamically Similar conditions, subjects walked at 4 levels of BWS: 0, 20, 40, and 60% of body weight. At 0% BWS, no support was provided. Walking speed for each BWS level was calculated from the criteria for dynamic similarity (Alexander, 1989). At all levels of BWS, subjects walked at Fr of 0.25, which corresponds to the preferred walking speed (Alexander, 1989; Leurs et al., 2011), and was calculated as:

$$v = \sqrt{Fr * g * L}$$

where v is the walking velocity (m/s), g is the scaled gravitational constant (m/s^2), and L is leg length (m). The gravitational constant was scaled depending upon the level of BWS provided.

Speed-Matched Conditions

During the Speed-Matched conditions, subjects walked at three levels of BWS: 20, 40 and 60% of body weight. At all levels of BWS, walking speed was the same as the 0% BWS at Fr = 0.25 condition.

Analysis

Data analysis

Kinematic marker data was processed in Visual3D (C-Motion, Germantown, MD). Marker data was gap-filled and low-pass filtered (Butterworth, cut-off frequency 6 Hz). Gait events (initial contact and toe-off) were identified by the fore-aft position of the calcaneal and 5th metatarsal markers: initial contact was identified as the local maximum of the calcaneal marker per step and toe-off as the local minimum of the 5th metatarsal per step. All trials were visually inspected to verify accurate event detection.

The processed marker data and gait events were used to calculate the following metrics for each step: lateral restoring force, step width, step width variability, step length, step length variability, and minimum lateral margin of stability (MOS). We estimated the peak lateral restoring force

$F_{lateral}$ (N), as a function of CoM excursion and vertical BWS force, the equation is included as supplementary material.

Step width and length were calculated as the medio-lateral and fore-aft distance, respectively, between the left and right calcaneus markers at initial contact. These variables were averaged across all steps for each subject and condition. Similarly, variability was calculated as the standard deviation of these measures.

We calculated the minimum lateral MOS (m) for each step using the procedure from Hof et al. (2005), details of which are included in the supplementary material. To account for the effect of BWS, the gravitational constant in the MoS equation was scaled to the level of BWS used during the testing condition. Minimum lateral MoS was then calculated during each stride, the minimum found, and then averaged over all strides for each subject and condition.

Statistical analysis

Statistical analysis was performed in SPSS (IBM, Armonk, NY). Data were compared using two one-way repeated measure analysis of variance (ANOVA) to test for differences in each metric. Dynamically Similar and Speed-Matched conditions were separated with the independent variable BWS (levels of 0, 20, 40, and 60%). When a significant main effect of BWS was found, Dunnett’s test for multiple comparisons was performed to look for differences between individual BWS levels and the control condition (0% BWS). For the lateral restoring force metric, where there was no control condition, comparisons were made between all groups and a Tukey-Kramer test was used. For all statistical tests, a significance level of $p < 0.05$ was used. If sphericity was violated, the Greenhouse-Geisser F-statistic and p-value were used to test the main effects.

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Results

Trajectories, Walking Speeds / Froude Numbers, and Lateral Restoring Forces.

For the Dynamically Similar conditions, the walking velocities at each BWS level were 1.47 ± 0.08 m/s at 0%, 1.31 ± 0.06 m/s at 20%, 1.13 ± 0.05 m/s at 40%, and 0.93 ± 0.04 m/s at 60% BWS. For the Speed-Matched conditions, a walking velocity of 1.47 ± 0.08 m/s was used at all BWS levels, which corresponded to a mean Fr of 0.31, 0.42, and 0.63 at BWS of 20%, 40%, and 60%, respectively.

Peak lateral restoring force showed a significant main effect of BWS for both the Dynamically Similar (ANOVA; $p < 0.001$) and Speed-Matched (ANOVA; $p < 0.001$) conditions. (Figure 2.2B). For both Dynamically Similar and Speed-Matched conditions, each BWS level was significantly different than the other two (Tukey-Kramer; $p < 0.05$) and increased from 20% to 60%.

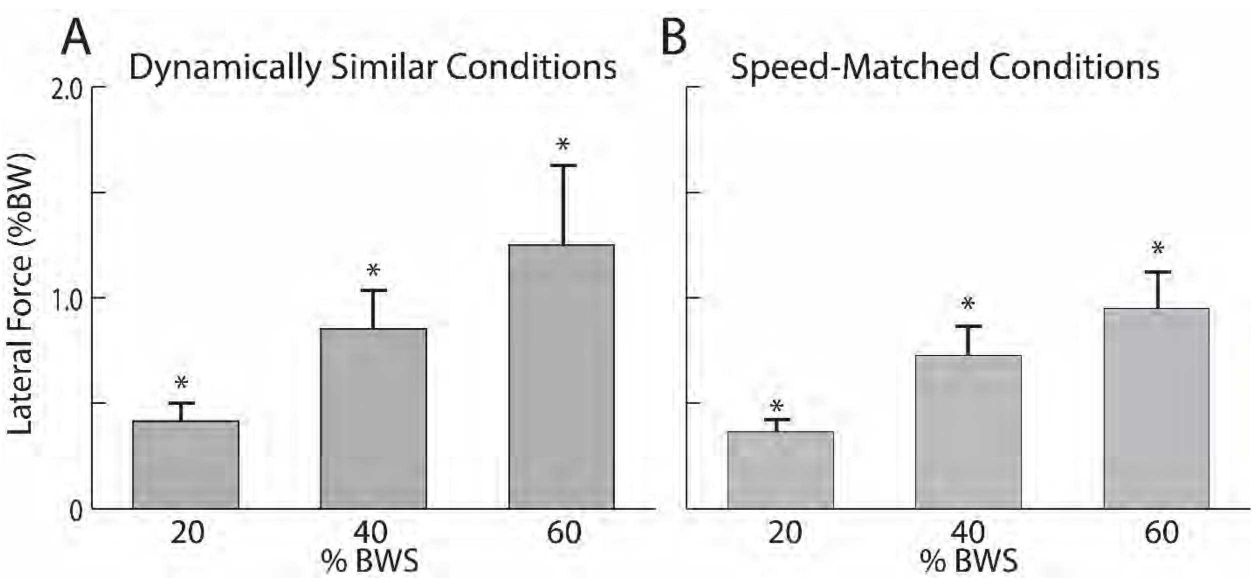

Figure 2.2: Lateral Restoring Force

Step Width and Variability

Step width significantly increased for the Dynamically Similar (Figure 2.3, 2.4A) (ANOVA; $p = 0.037$) and Speed-Matched (Figure 2.3, 2.4B) (ANOVA; $p = 0.001$) conditions as BWS increased. In both Dynamically Similar and Speed-Matched conditions, 60% BWS step width was significantly wider than 0% BWS (Dunnett's; $p < 0.05$).

There was a significant reduction in step width variability for both the Dynamically Similar (Figure 2.4C) (ANOVA; $p < 0.001$) and Speed-Matched (Figure 2.4D) (ANOVA; $p < 0.001$)

conditions when BWS was provided. Pairwise comparisons found that step width variability for all three levels of BWS were significantly smaller than at 0% BWS (Dunnett's; $p < 0.05$) during both conditions.

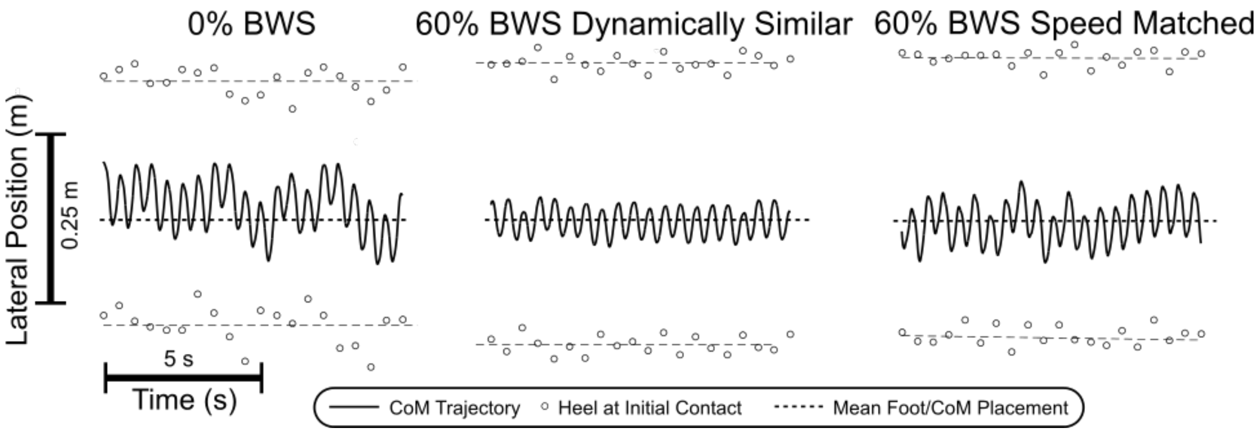

Figure 2.3 Effect of BWS on CoM and Foot Position Mean estimated lateral force provided by the BWS system for (A) Dynamically Similar and (B) Speed-Matched conditions. Conditions 20%, 40%, and 60% represent the amount of BWS provided. Significance ($p < 0.05$) is denoted by * and indicates each level is significantly different than the other two.

Step Length and Variability

Step length significantly decreased for the Dynamically Similar (Figure 2.5A) (ANOVA; $p = 0.002$) condition as BWS increased. Pairwise comparisons found that 60% and 40% BWS produced significantly shorter steps than 0% BWS for Dynamically Similar (Dunnett's; $p < 0.05$) conditions. There was no change in step length for the Speed-Matched condition (Figure 2.5B) (ANOVA; $p > 0.05$).

Step length variability showed a significant main effect of BWS for both Dynamically Similar (ANOVA; $p < 0.001$) and Speed-Matched (ANOVA; $p < 0.001$) conditions. Step length variability increased for both conditions (Figure 2.5C and D) at 60% BWS compared to 0% BWS (Dunnett's; $p < 0.05$).

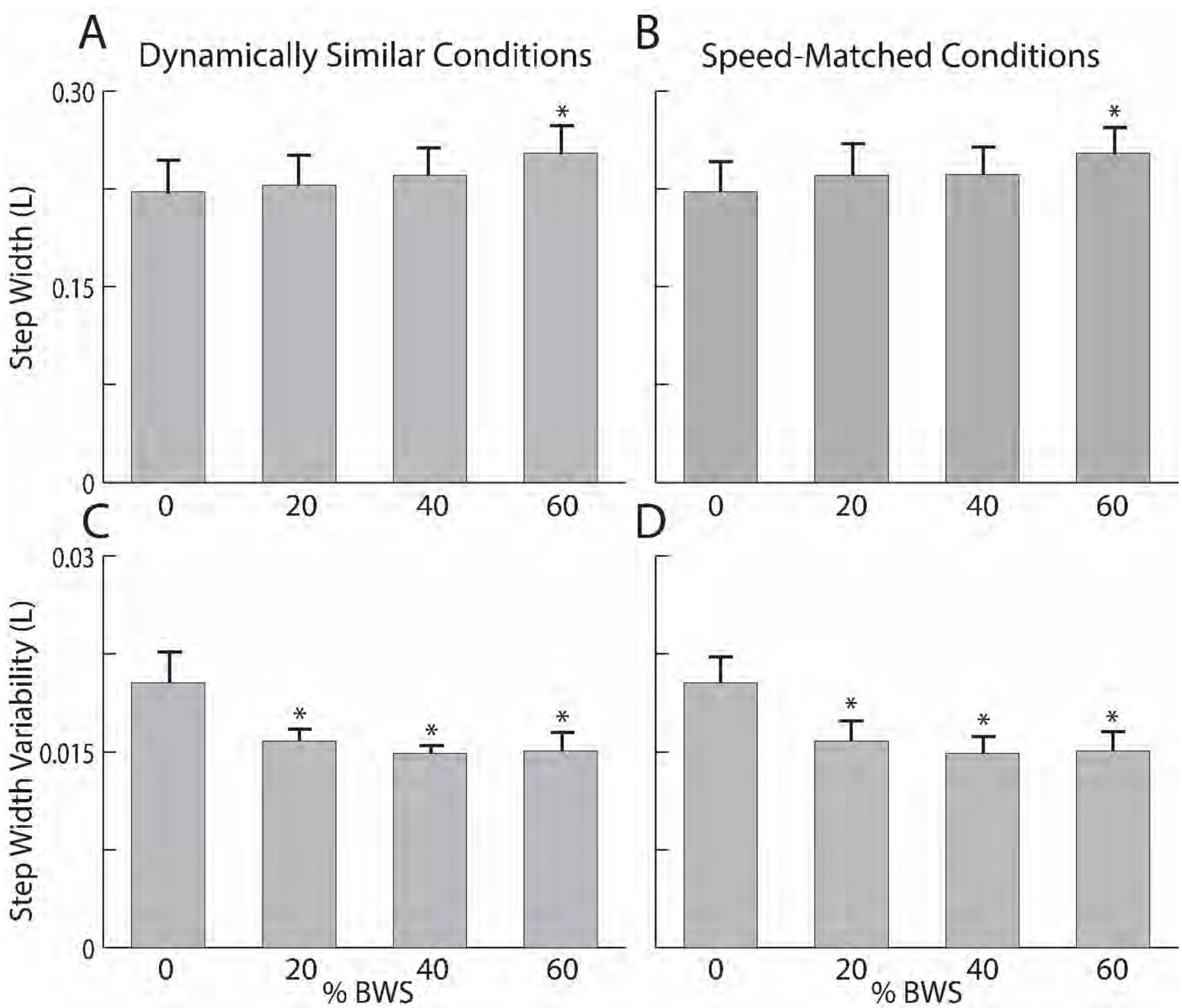

Figure 2.4 Effect of BWS on Step Width and Step Width Variability Mean + SD step width comparisons for (A) Dynamically Similar and (B) Speed-Matched conditions. (C) Mean + SD

step width variability comparisons for Dynamically Similar and (D) Speed-Matched conditions. Significance (p < 0.05) from 0% BWS is denoted by *.

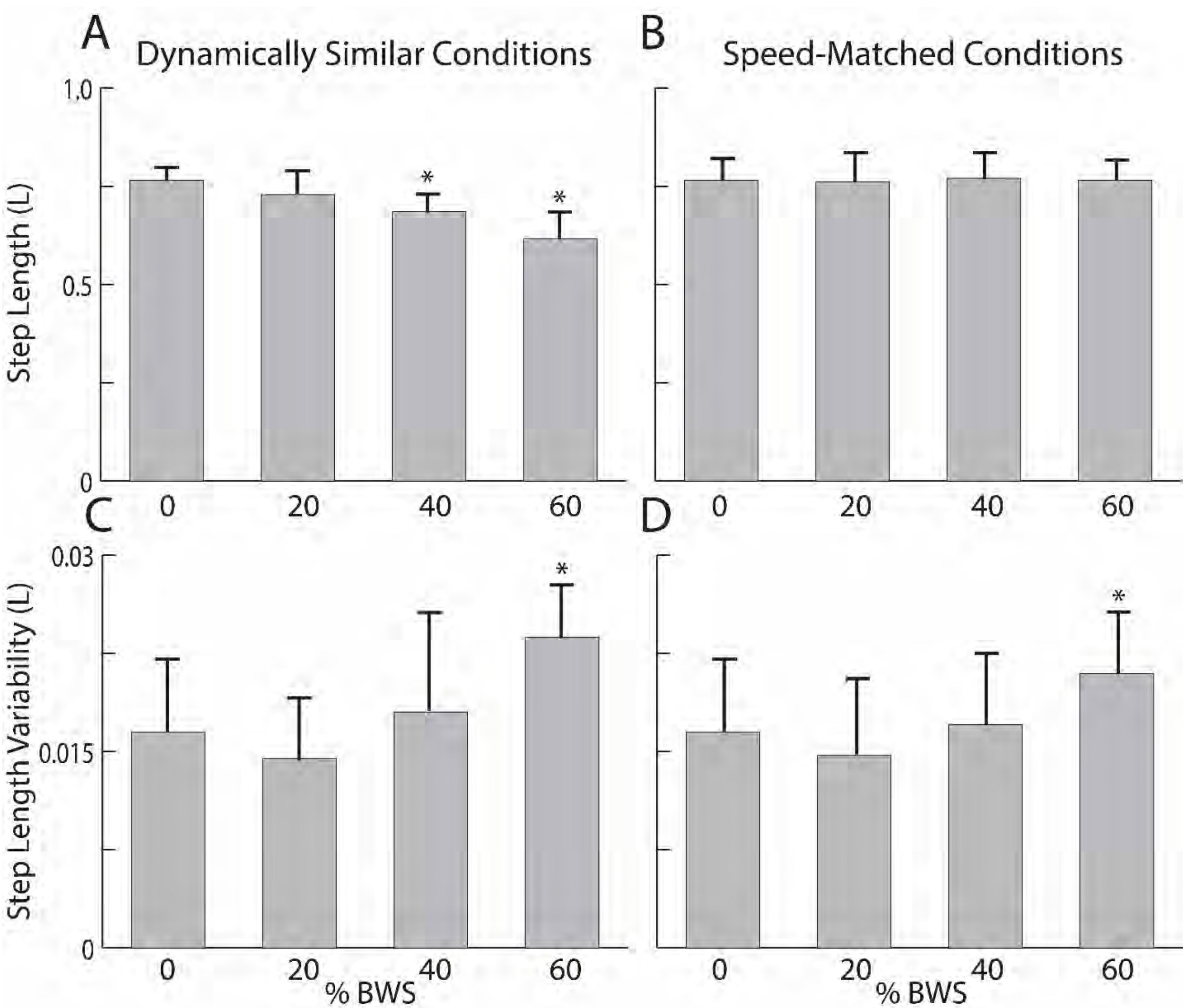

Figure 2.5 Effect of BWS on Step Length and Step Length Variability Mean + SD step length comparisons for (A) Dynamically Similar and (B) Speed-Matched conditions. Mean + SD step length variability comparisons for (C) Dynamically Similar and (D) Speed-Matched conditions. Significance (*p* < 0.05) from 0% BWS is denoted by *.

Lateral Margin of Stability

There were no significant differences in MOS across BWS levels for either the Dynamically Similar (Figure 2.6A) (ANOVA; p = 0.534) or Speed-Matched (Figure 2.6B) (ANOVA; p = 0.083) conditions.

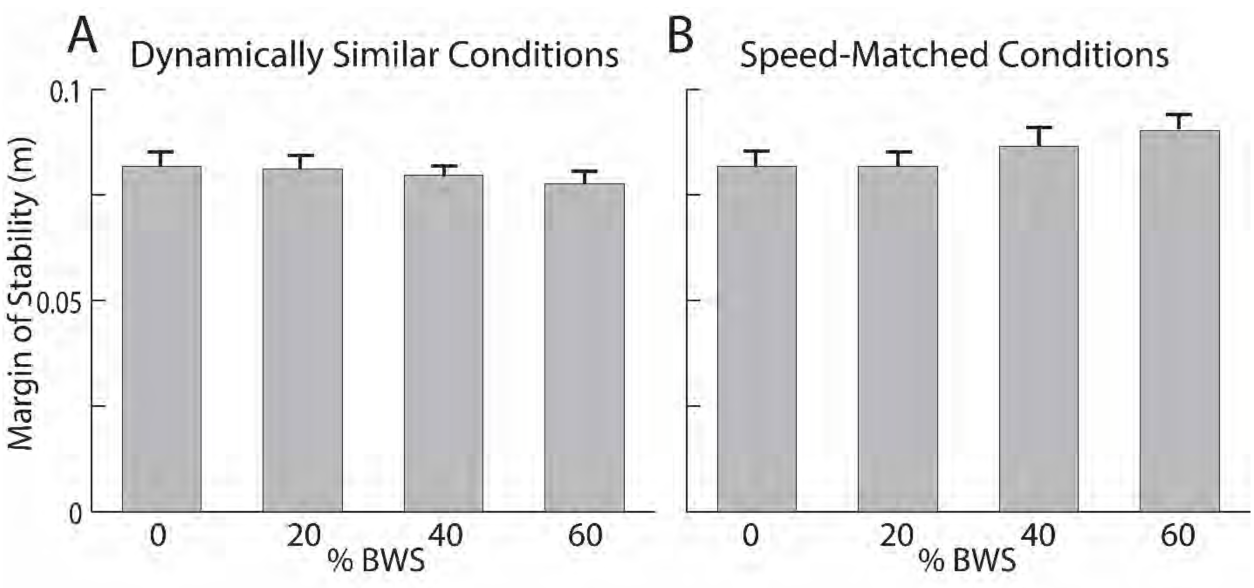

Figure 2.6 Effect of BWS on Minimum Lateral MOS Mean + SD for Minimum lateral Margin of Stability for (A) Dynamically Similar and (B) Speed-Matched conditions. There were no significant comparisons for this measure.

Discussion

Our purpose was to quantify the interaction between BWS and the requirements to actively stabilize in the frontal plane. We examined two walking conditions based on our assumption that the two major factors influencing lateral stability with BWS are 1) the presence of lateral restoring forces that would decrease the requirements of active lateral stabilization, and 2) reductions in stabilizing gravitational moments that would increase the requirements to actively control stability. To investigate these two effects, individuals walked with varying BWS levels in either Dynamically Similar or Speed-Matched conditions. We assumed that the effect of BWS on the ability of the gravitational moment to counteract lateral inertial moments would be consistent across the Dynamically Similar conditions.

Our findings were mixed. In both the Dynamically Similar and Speed-Matched conditions, individuals demonstrated significant reductions in step width variability at all BWS levels when compared to 0% BWS. However, the reduction in step width variability with BWS was not accompanied by decreases in step width or lateral MOS. In both the Dynamically Similar and Speed-Matched conditions, step width actually increased at the highest BWS level. That step width variability decreased while step width was either unchanged or increased with increasing BWS levels suggests that the effects of BWS on the requirements of lateral stability are not straightforward.

The reduction of step width variability was the strongest evidence that BWS reduced the requirements of lateral stability. This decrease in variability was consistent with previous research documenting increased medio-lateral trunk regularity with BWS in both able-bodied [48] and hemiplegic [76] gait. These decreases in variability could be attributable to external lateral restoring forces created by the BWS system. Similarly, able-bodied subjects have been shown to reduce step width variability by 33% [36] when given direct lateral external stabilization. This is comparable to the ~25% reductions in step width variability we observed in the current study when subjects received some BWS. The reduced variability with external lateral stabilization has been suggested to indicate a decrease in demands on the nervous system to actively control medio-lateral foot placement [36].

However, changes in variability are difficult to interpret. Step width variability could be reduced by either decreases in the mechanical requirements of lateral stability (i.e. the presence of external lateral stabilization) or by increases in a person's ability to control foot placement. As such, it is possible that the reduction in step width variability was not due to BWS reducing the mechanical requirements of lateral stability. The magnitude of the lateral restoring forces we observed with BWS were fairly small, ranging from 0.36 to 1.25% body weight. While these values were similar in magnitude to those reported previously [75], lateral restoring forces less than 0.5% body weight may not be sufficient to noticeably decrease the mechanical demands of lateral stabilization. One limitation of this study is that we calculated the restoring forces based on an assumed rather than measured constant support force provided by the BWS system. However, errors in total BWS force created by the support system used in the current study have been previously found to be less than 1 kg during human walking [47].

Given that step width variability did not decrease proportionally as lateral restoring forces increased, the relationship between these variables may not be entirely mechanical. It is possible that the major effect of the laterally directed force on mediolateral variability was to provide additional sensory information on subjects' lateral position on the treadmill. BWS may have acted as an external reference point that improved subjects' ability to control their limb trajectory. This increased position sense could have resulted in reduced step width variability and the appearance of reduced mechanical requirements for lateral stability. Indeed, other research has suggested that light touch on a hand rail during treadmill walking can significantly reduce gait variability [79]. In support of this idea, step length variability, unlike step width variability, actually increased with BWS. The BWS system used in the current study actively tracked subject's fore-aft position while resisting medio-lateral translations. Thus, the BWS forces would have provided subjects with additional haptic feedback about medio-lateral but not fore-aft position. This difference in feedback could explain the difference in changes between step width and step length variability with BWS.

In contrast to step width variability, the changes we observed in step width and lateral MoS suggest that increasing BWS may actually increase lateral stability requirements. Previous research has found that people select narrower steps and a smaller lateral MOS, when given external lateral stabilization during treadmill walking [34], [36], [80] and tend to take wider steps and increase lateral MoS when placed in unstable environments [81], [82]. Consequently, if BWS were to increase the requirements to laterally stabilize, then we would expect step width and lateral MoS to increase with BWS. That BWS may be destabilizing is supported by previous research demonstrating increases in sagittal plane variability [52], [57] as well as differences in temporal-spatial and kinematic variables of able-bodied gait with high levels of BWS [58], [83], [84]. However, we observed a decrease in variability with BWS. Past research has not found a relationship between step width and step width variability [37] it is possible that these two measures are not independent. MOS tended to increase with BWS in the Speed-Matched condition, however this increase was not significant and may have been due to our small sample size (n = 8). In addition, step length only reduced when walking speed was decreased in the

Dynamically Similar condition, but not in the Speed-Matched condition, which is consistent with other external lateral stabilization studies [34], [85].

At 60% BWS in the Speed-Matched condition, subjects walked at Fr of 0.63. This value is insightful because humans typically transition from walking to running at a Fr greater than ~0.5 [10], [86], [87]. One possible explanation why humans transition from walking to running is that these respective modes of locomotion will improve gait stability [88] and minimize energetics cost [89] at the associated gait speeds. In the current experiment, dictating walking in conditions where running should have been the preferred gait pattern may have led to increases in lateral stability requirements.

In contrast, for the Dynamically Similar condition, we hypothesized that there would be a decrease in step width and lateral MOS with BWS due to the lateral restoring force. However, for the Dynamically Similar condition we found an increase in step width at 60% BWS and no change in lateral MOS. We assumed that the ratio of inertial to gravitational forces would be consistent in the Dynamically Similar conditions by changing absolute walking speed. However, changing absolute walking speed may not influence frontal and sagittal plane inertial forces proportionally. As such the Dynamically Similar conditions may not have been purely examining the effects of the lateral restoring forces. Our results appear to disagree with a previous study that had observed a decrease in step width as BWS increased [46]. A key difference between these studies was that mass was added to the subjects' bodies to offset the BWS force in the previous work. The added inertia could have independently affected lateral stability metrics.

This study has several important clinical implications. If the goal of gait training is to improve control of frontal plane stability, it is important to realize that BWS may provide lateral restoring forces and supplemental sensory information that make it easier for individuals to control lateral stability. If stability is to be challenged, obtaining a BWS system that eliminates these lateral restoring forces could be beneficial [74], [75]. However, clinical populations often receive other direct methods of stabilization (e.g., holding on to hand rails) that likely have larger lateral stabilization effects than BWS. If a primary goal of gait rehabilitation is to increase speed, then lower levels of BWS should be used to avoid situations where Fr is greater than 0.5 and running becomes the preferred mode of locomotion. Finally, it is important to understand that high levels of BWS will likely induce complex changes in the body's passive walking dynamics that could create control challenges in multiple planes of motion. Walking at high speeds with high levels of BWS may create a walking environment that does not fully replicate the active control requirements needed for real world walking. Further work examining the stability implications of BWS on impaired populations are necessary to fully evaluate the clinical implications.

Acknowledgments

This work was supported in part by Career Development Award #1 IK2 RX000717-01 from the United States Department of Veterans Affairs, Rehabilitation Research and Development Service and NIH training grant T32HD057845. We would also like to thank Mary Wu for her assistance analyzing the data.

Supplementary Material

The equation that we used to calculated peak lateral restoring force was taken in modified form from Figure 1 and Equation 23 in [90]. Our equation for the lateral restoring force F_l (N):

$$x_p = \frac{COM_E}{2}$$

$$\sin(\theta) = \frac{x_p}{\sqrt{{x_p}^2 + h^2}}$$

$$F_l = F_{BWS}\sin(\theta)$$

Where x_p is the maximum deviation of the COM from the midline of the BWS suspension (m), COM_E is the lateral excursion of the COM (m), Θ is the angle of the support cable with respect to the vertical axis, h is the distance between the BWS system and the height of the shoulder (m) [91], F_{BWS} is the force provided by the BWS system (N). The maximum deviation of the COM was not calculated directly. We made the assumption that the midline of the BWS suspension was equivalent to the mean mediolateral COM position. This lateral restoring force was then normalized by body weight and averaged for each subject and the six BWS conditions.

Equation for minimum lateral MOS (m) modified from [13]:

$$XCOM = COM_p + COM_v\sqrt{\frac{l}{g}}$$

$$MOS = XCOM - BOS$$

Where $XCOM$ is the extrapolated center of mass position in the frontal plane (m), COM_p is lateral center of mass position (m), COM_v is the lateral center of mass velocity (m/s), and BOS is the lateral base of support position (m). The lateral position of the center of mass was estimated as the midpoint of the greater trochanter markers. The center of mass velocity was calculated as the first derivative of the center of mass position and the BOS was estimated from the lateral position of the 5th metatarsal markers. Minimum lateral MOS was then calculated for each step and averaged over all steps for each subject during each condition. Our examination of minimum MOS was restricted to the frontal plane.

Chapter 3: Walking with Body-Weight Support Reduces Coordination between Lateral Foot Placement and Center of Mass Motion

Abstract

Background: Body-weight support (BWS) systems are a common clinical tool for gait and balance training. BWS alters walking dynamics and may encourage lateral stepping strategies that are non-optimal for controlling center of mass (COM) motion during real-world walking.

Research Question: Does BWS alter frontal-plane stepping coordination during gait?

Methods: We measured lower extremity stepping behavior as 9 healthy young adults walked on a treadmill with different levels of BWS. We characterized the effects of BWS on COM motion and calculated coordination between the COM state (position and velocity) during swing phase and the subsequent lateral foot placement. We compared stepping behavior and coordination between levels of BWS and between limbs.

Results: BWS significantly reduced coordination between COM state and lateral foot placement ($p < 0.001$, $F(2,16) = 6.528$) and reduced symmetry of the COM trajectory in the vertical and frontal planes ($p < 0.001$, $p < 0.001$). As BWS increased, there was increased bias of the COM trajectory towards the right side of the body. There was a significant interaction between BWS and limb on lateral foot placement ($p < 0.001$), and as BWS increased the right foot was placed more lateral to the COM than the left foot.

Significance: Our results suggest that young adults adopt a compensatory stepping strategy when walking with BWS where there is reduced coordination between the COM and the swing foot. This stepping strategy may have limited translation to real-world walking dynamics and BWS may discourage active coordination of frontal-plane stepping.

Introduction

Body-weight support (BWS) systems are a common rehabilitative tool that has been used successfully to improve walking function of people with neuromusculoskeletal deficits [45], [71], [72], [92]. Although BWS treadmill training has not been found to be more effective than overground interventions [9], [93], this setup is often desirable in clinical settings because it creates a safe environment to practice many steps and is easy to progressively challenge patients [93]–[95]. However, clinicians may need to use discretion when using BWS if a primary goal is to retrain walking balance. The concern is that the novel center of mass (COM) dynamics created when using BWS may promote practice of stepping behaviors that are non-optimal for controlling unassisted walking [1].

Simple biomechanical models suggest that walking is passively unstable in the frontal-plane and requires active control [23]. BWS may interfere with learning walking balance by disrupting frontal-plane COM dynamics in several ways. BWS will counteract the gravitational moment acting about the subtalar joint during single limb support that plays a primary role in controlling the whole body lateral inertial moment [1], [11]. In addition, if BWS is provided through a fixed overhead support it will apply a lateral force to the body when the user is not directly below the system [1], [46], [74]. Finally, offsetting gravity will alter the spatiotemporal characteristics of the

COM inverted-pendulum trajectory [96]. The disruption to normal frontal-plane walking dynamics should increase in proportion to the level of BWS and may encourage compensatory stepping strategies.

Modulating lateral foot placement is a primary method to control frontal-plane walking dynamics. High levels of BWS consistently result in changes in lateral foot placement during walking. However, there is not a consensus as to how and why foot placement changes with BWS. In response to high levels of BWS studies have reported that people take both narrower [46] and wider [1] steps. Similarly, gait variability has been reported to both increase [52], [57] and decrease [1] in response to BWS. A challenge of interpreting these changes in lateral foot placement is that without information about the corresponding COM dynamics, it is not evident how these modulations contribute to control of frontal-plane stability.

During normal walking there is a correlation between step-to-step lateral foot placement and lateral COM state (e.g., on steps when the COM is positioned farther to the right, people take wider steps to the right) [17], [97]–[99]. Of note, the strength of this correlation scales (decreasing in stabilizing environments and increasing in destabilizing environments) to the demands of controlling frontal-plane dynamics [63], [66]. While BWS may change frontal-plane dynamics, it is not evident that this external force will disrupt the relationship between foot placement and COM state as certain forms of lateral stabilization can be used without disrupting this coordination [97]. Thus, it is important to understand how BWS impacts the step-to-step control of frontal-plane walking dynamics.

Therefore, our purpose was to investigate the effects of BWS on the control of frontal-plane walking dynamics by quantifying the step-to-step relationship between COM state and foot placement. We hypothesized that increasing levels of BWS would reduce the correlation between lateral COM state and lateral foot placement.

Methods

Participants

The Northwestern University Institutional Review Board approved the study. Nine healthy young adults (4 female, 25 ± 2 years old, 78 ± 15 kg) provided written, informed consent. Participants were all able to walk continuously for 10 minutes, had no known musculoskeletal or neurological conditions affecting gait or balance, and were not taking medications affecting gait or balance.

Experimental Setup

Participants performed a series of walking trials on an instrumented, split-belt treadmill (Motek Medical B.V., Houten, Netherlands). Participants wore a trunk harness attached to an overhead body-weight support system (Aretech, Ashburn, VA) with an actuated trolley oriented perpendicular to the treadmill belt, allowing for natural lateral oscillation during walking.

We used a 12-camera motion capture system (Qualisys, Gothenburg, Sweden) to record the 3D positions of 38 retro-reflective markers at 100 Hz. Markers were affixed to the sternum and bilaterally on the acromion, anterior and posterior iliac spine, greater trochanter, lateral epicondyle of the knee, lateral malleolus, calcaneus, and the heads of the 2nd and 5th metatarsals. Additional clusters of 4 markers were affixed bilaterally to the thigh and shank. Surface EMG electrodes (Delsys, Natick, MA) were affixed to the belly of the gluteus medius muscle after the skin was prepared according to SENIAM guidelines [100] and recorded at 1000 Hz.

Protocol

Participants performed walking trials at their preferred speed at three unloading conditions: 0%, 30%, and 60% BWS. At 0% BWS, participants were not attached to the overhead system but did wear the trunk harness. Participants walked for three minutes per trial and repeated each condition twice. The first minute of each trial allowed participants to accommodate to the condition. The trial order was randomized, and participants were allowed to rest between trials.

Data Analysis

Kinematic and kinetic data were processed with Qualisys Track Manager, Visual3D (C-Motion, Germantown, MD) and custom MATLAB (Mathworks, Natick, MA) programs. Marker and force plate data were low-pass filtered (4^{th}-order Butterworth, 6 Hz cut-off frequency) and gap-filled. Gait events were detected by a 5 N threshold on the vertical ground reaction force vector. The mediolateral position and velocity of the COM were calculated from the center of a pelvis segment [101].

To compare the effects of BWS on COM state, we created a set of Lissajous curves which represents COM position as a series of parametric equations and can be used to compare the cyclical signals of the gait cycle. This method [102] creates a convoluted loop of the 3D COM trajectory that retains time as an independent variable. For each participant and condition, we fit the x-, y-, and z-position of the COM during the gait cycle to a Fourier Series:

$$t = 2\pi \frac{t'}{T}$$

$$\hat{x}(t) = \sum_{i=1}^{6} c_i^x \sin{(it + \varphi_i^x)}$$

$$\hat{y}(t) = \sum_{i=1}^{6} c_i^y \sin{\left(it + \varphi_i^y\right)}$$

$$\hat{z}(t) = \sum_{i=1}^{6} c_i^z \sin{(it + \varphi_i^z)}$$

Where t' and t are the absolute and normalized time respectively, T is the stride duration, i is the harmonic number, c_i is the harmonic single-sine coefficient, and φ_i is the harmonic single-sine phase. This procedure was repeated for each subject and experimental condition, and then a mean COM trajectory was computed for each BWS condition across participants.

Following the Fourier Analysis, we computed a symmetry index for the frontal-plane (x-z plane) $S^x$and the vertical-plane (x-y) S^y :

$$S^x = \frac{c_2^x + c_4^x + c_6^x}{\sum_{i=1}^{6} c_i^x} \quad S^y = \frac{c_2^y + c_4^y + c_6^y}{\sum_{i=1}^{6} c_i^y}$$

These symmetry indexes range from 0 to 1. A symmetry of 1 indicates that there is perfect symmetry between the left and right sides.

To quantify stepping coordination, we used a regression equation to predict the next mediolateral foot placement position based on the COM state (position and velocity) during the preceding swing phase. This method [17] calculates a time-series coefficient of determination (R^2), which is the ratio between the variance in lateral foot placement position and the variance in COM state. As R^2 increases, there is a smaller difference between predicted and actual foot placement position, and a stronger correlation emerges. We used the following regression equation to measure stepping coordination:

$$FP_x = \beta_1(i) \cdot COM_x(i) + \beta_2(i) \cdot COM_v(i) + \varepsilon(i)$$

where FP_x is lateral foot placement position relative to the contralateral stance foot, COM_x is the lateral COM position, COM_v is the lateral COM velocity, $\beta_1$and β_2 are regression coefficients, ε is the error, and i is a discrete time point during the preceding swing phase. This process was repeated every 2% of the swing phase, resulting in a time-series R^2 value.

We calculated step width and its variability, right and left foot placements, COM excursion, maximum COM velocity, peak hip abduction angle during the swing phase, peak hip adduction angle during the stance phase, and frontal-plane hip ROM as additional outcome measures. Step width was calculated as the mediolateral distance between the 5^{th} metatarsals at each heel strike. Right and left foot placements were calculated as the mediolateral distances between the COM and the right and left 5^{th} metatarsals, respectively, at heel strike (Figure 3.3). For both step width and foot placement, values were normalized to the percentage of a participant's leg length. For each measure, only the last 100 steps of each trial were analyzed.

Gluteus medius EMG data were detrended, rectified, and filtered using a moving-average filter. The maximum value of gluteus medius activity during the 0% BWS condition was used to normalize the data across conditions. We then calculated the root mean square (RMS) value during time windows in the stance and swing phases. During the stance phase of gait, the gluteus medius acts to stabilize the pelvis and assist with medial redirection of the COM, so we calculated RMS EMG from heel strike to contralateral toe off (0-40% of the gait cycle) [11]. We calculated gluteus medius RMS EMG during the first half of swing phase (60-80% of the gait cycle) as the gluteus medius acts to redirect the swing limb in the frontal-plane and is responsible for active neuromuscular control of foot placement during this period [22].

Statistical Analysis

We used MATLAB and SPSS (IBM, Armonk, NY) for all statistical analysis. To compare stepping coordination across the swing phase, we used a statistical parametric (SPM) approach [103]. We first performed a SPM paired t-test of R^2 time-series between legs at each condition. If there were no significant differences between legs, a R^2 time-series was calculated with data collapsed from both legs within each BWS condition.

To compare stepping coordination between conditions, we performed a one-way repeated measures ANOVA with a fixed effect of BWS condition (0%, 30%, and 60%). This approach produces a F-value for each sample of the R^2 time-series. When F-values cross a set threshold value, corresponding to $\alpha = 0.05$, it indicates a statistically significant effect. If there was a significant main effect of BWS, Bonferroni-corrected pairwise comparisons were then analyzed. For outcome measures that were bilateral we fit separate linear mixed effects model in SPSS with fixed effects of BWS, leg, and the interaction between BWS and leg, and random intercepts allowed each participant to deviate from the main intercept. If there was a significant main effect, we performed Bonferroni-corrected comparisons. If there was a significant interaction between main effects, we performed leg comparisons within each BWS condition (e.g., right vs left at 0% BWS). For foot placement, we also performed BWS comparisons within each leg (e.g., 0% BWS vs 30% BWS within the right leg). Significance was set to $p < 0.05$ for all statistical tests. To compare the additional outcome metrics, we created separate linear mixed effects models in SPSS with fixed effects of BWS and random intercepts allowed each participant to deviate from the main intercept. Bonferroni-corrected comparisons were calculated if there was a significant effect of BWS.

Table 3.1 Results of Linear Mixed Effects Models

Metric	Fixed Effect (p-value)	Post-hoc (Corrected p-value)
Step width	BWS (0.01)	0 > 30 (0.31)
		0 < 60 (0.26)
		30 < 60 (<0.001)
Step width variability	BWS (0.001)	**0 > 30 (<0.001)**
		0 > 60 (0.16)
		30 < 60 (0.10)
ML COM Excursion	BWS (<0.001)	**0 > 30 (<0.001)**
		0 > 60 (<0.001)
		30 > 60 (<0.001)
AP COM Excursion	BWS (<0.001)	**0 > 30 (<0.001)**
		0 > 60 (<0.001)
		30 > 60 (<0.001)
Maximum ML COM Velocity	BWS (<0.001)	**0 > 30 (< 0.001)**
		0 > 60 (<0.001)
		30 > 60 (<0.001)
Maximum AP COM Velocity	BWS (<0.001)	**0 > 30 (<0.001)**
		0 > 60 (<0.001)
		30 > 60 (<0.001)
S^x	BWS (<0.001)	0 > 30 (1.000)
		0 > 60 (<0.001)
		30 > 60 (<0.001)
S^y	BWS (<0.001)	0 > 30 (1.00)
		0 > 60 (<0.001)
		30 > 60 (<0.001)

Metric	Fixed Effect (p-value)	Post-hoc (Corrected p-value)
Foot placement	BWS (0.04)	0 > 30 (0.65)
		0 < 60 (0.57)
		30 < 60 (0.04)
	Leg (<0.001)	**Right > Left (<0.001)**
	BWS*Leg (<0.001)	0: Right > Left (0.01)
		30: Right > Left (<0.001)
		60: Right > Left (<0.001)
		Right: 0 < 30 (0.01)
		Right: 0 < 60 (<0.001)
		Right: 30 < 60 (<0.001)
		Left: 0 > 30 (<0.001)
		Left: 0 > 60 (<0.001)
		Left: 30 > 60 (0.06)
Peak Swing Phase Hip Abduction	BWS (0.11)	
	Leg (<0.001)	**Right > Left (<0.001)**
	BWS*Leg (<0.001)	0: Right > Left (0.30)
		30: Right > Left (<0.001)
		60: Right > Left (<0.001)
Peak Stance Phase Hip Adduction Angle	BWS (<0.001)	0 > 30 (1.00)
		0 < 60 (<0.001)
		30 < 60 (<0.001)
	Leg (<0.001)	**Right > Left (<0.001)**
	BWS*Leg (<0.001)	0: Right > Left (0.11)
		30: Right > Left (<0.001)
		60: Right > Left (<0.001)
Frontal Plane Hip ROM	BWS (<0.001)	0 > 30 (0.92)
		0 > 60 (<0.001)
		30 > 60 (<0.001)
	Leg (0.001)	**Right < Left (<0.001)**
	B*L (0.21)	
Stance Phase Gluteus Medius RMS	BWS (<0.001)	**0 > 30 (0.01)**
		0 > 60 (<0.001)
		30 > 60 (<0.001)
	Leg (0.78)	
	B*L (0.76)	

Results

COM Excursion, Velocity, and Symmetry

To visualize the effects of BWS on COM state and to measure the symmetry of the COM trajectory, we created a Lissajous curve of COM position for each BWS condition (Figure 3.1) and evaluated the effects of BWS on measured symmetry indexes, COM excursion, and maximum COM velocity (Table 3.1). As BWS increased, COM excursion became increasingly smaller in the anteroposterior and mediolateral directions ($p < 0.001$) and maximum COM velocity became increasingly slower in the same directions ($p < 0.001$). In addition, as BWS increased, frontal and vertical plane symmetry of the COM became increasingly more asymmetric ($p < 0.001$).

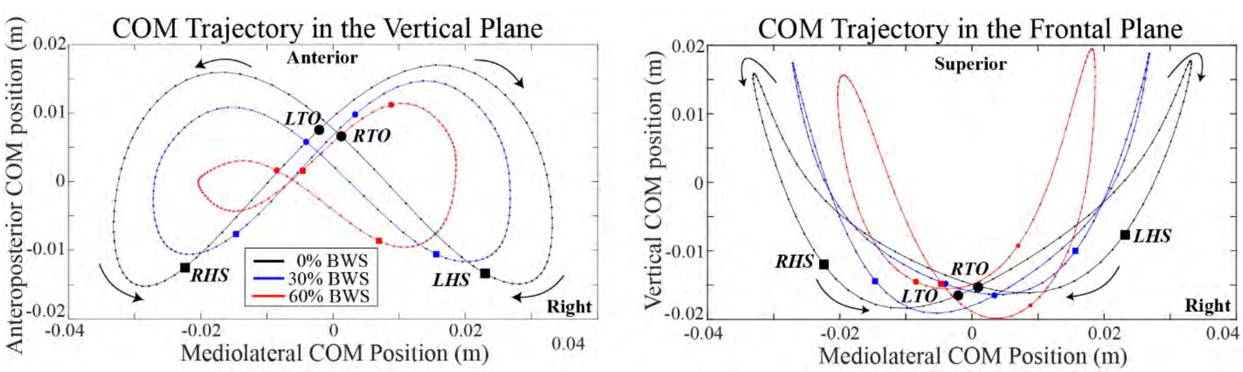

Figure 3.1 Effect of BWS on COM Trajectory in Vertical and Frontal-Planes Left: COM trajectory in the vertical-plane (x-y) for each level of BWS (0%, 30%, and 60%). On each trajectory, the distance between two dots represents 1% of the gait cycle. Filled squares and circles represent gait events (square: heel strike, circles: toe off). Arrows indicate the direction of the COM trajectory through each plane.

Stepping Coordination

There were no significant differences in R^2 time-series between legs at each level of BWS ($p > 0.05$), so the data were collapsed across limbs to calculate a single R^2 time-series for each condition. We found that as BWS increased, the correlation between lateral COM state and lateral foot placement decreased (Figure 3.2). There was a significant main effect of condition ($p < 0.001$, $F(2,16) = 6.528$) for the entirety of the swing phase and significant differences between all comparisons (Figure 3.3, 0% > 30%, $p < 0.001$, $t(1,34) = 3.089$; 30% > 60%, $p < 0.001$, $t(1,34) = 3.071$; 30% > 60%, $p < 0.001$, $t(1,34) = 3.152$).

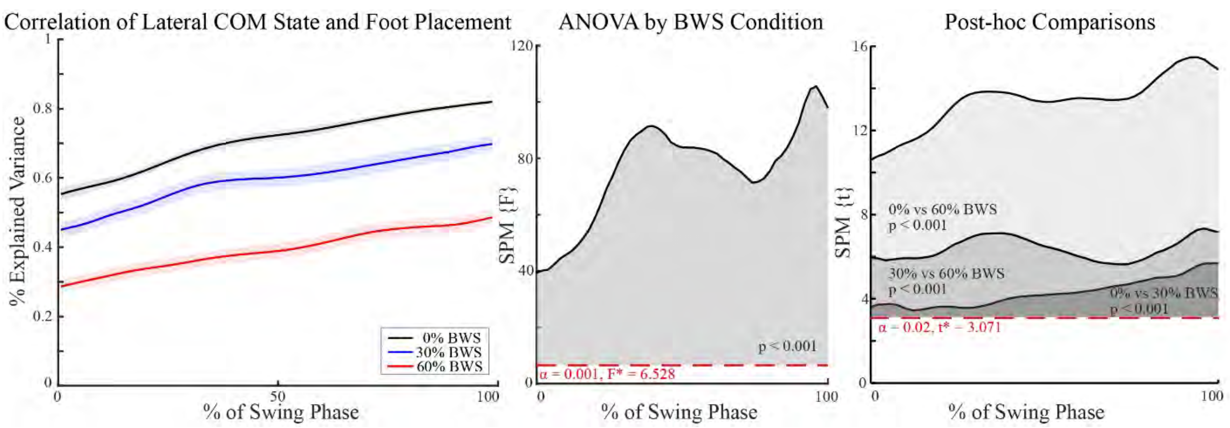

Figure 3.2 Effect of BWS on Correlation between Lateral COM State and Lateral Foot Placement Left: Mean ± standard error (shaded region) of R^2 time-series showing the capacity of COM state (position and velocity) to predict subsequent lateral foot placement during the swing-phase for each level of BWS (0%, 30%, and 60%). Center: Results from 1-way ANOVA with main effect of BWS on R^2 time-series data. The dashed red line indicates the F-threshold, corresponding to an $\alpha = 0.05$, where values above this threshold are statistically significant. Shaded regions indicate significant effects for the corresponding portion of the swing phase. Right: Differences in R^2 time-series between levels of BWS in order of effect size (top: 0% vs

60%, middle: 30% vs 60%, bottom: 0% vs 30%). The dashed red line represents the threshold t-value, corresponding to a post-hoc corrected $\alpha = 0.02$, where values above this threshold are statistically significant.

Step Width

For step width, we found a significant main effect of BWS (Figure 3.4, Table 1, $p = 0.006$) and that step width at 60% was larger than at 30% BWS ($p = 0.004$). For foot placement, there was a significant main effect of BWS ($p = 0.04$), leg ($p < 0.001$), and the interaction between BWS and leg ($p < 0.001$). As BWS increased, right foot placement became increasingly larger and left foot placement became increasingly smaller. For step width variability, we found a significant main effect of BWS ($p = 0.001$) and that step width variability at 0% BWS was greater than at 30% BWS ($p < 0.001$).

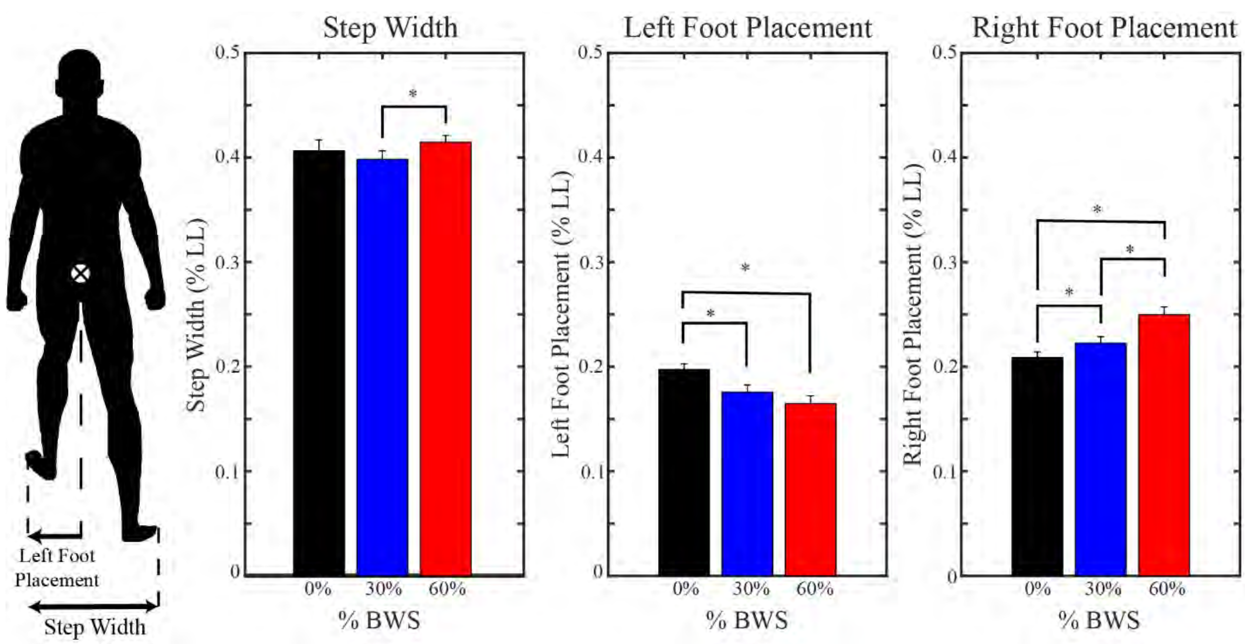

Figure 3.3 Effect of BWS on Lateral Foot Placement The diagram shows the differences between step width, the mediolateral distance between 5th metatarsals at heel strike, and foot placement, the mediolateral distance between the lateral COM position and 5th metatarsal at heel strike. Left: Mean step width ± standard effort for each levels of BWS (0%, 30%, and 60%). Center: Mean left foot placement ± standard error. Right: Mean right foot placement ± standard error. * indicates significant comparisons between BWS conditions (see Table 1 for further details).

Frontal-Plane Hip Kinematics

As BWS increased, we found that the right hip became more abducted during the stance phase (Figure 3.4, $p < 0.001$) and that the left hip became more adducted during the swing phase ($p < 0.001$). Frontal-plane excursion decreased as BWS increased ($p < 0.001$) and excursion on the right leg was reduced compared to the left leg ($p = 0.001$).

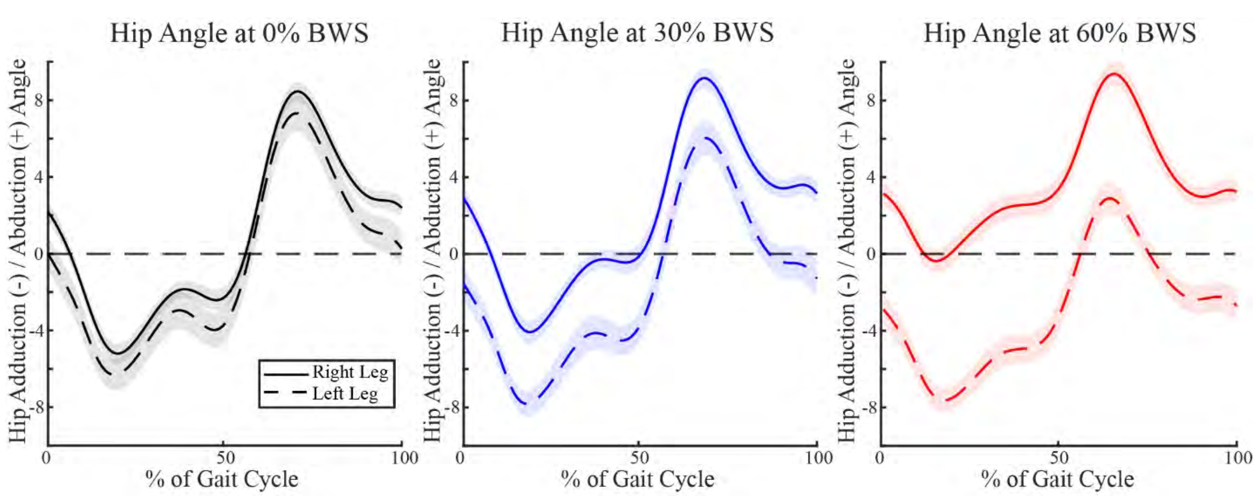

Figure 3.4 Effect of BWS on Frontal-Plane Hip Angles Mean ± standard error of frontal-plane hip angles at 0% (left), 30% (center), and 60% BWS (right) for the right (solid line) and left (dashed line) legs. See Table 1 for details of statistical tests.

EMG Activity of Gluteus Medius

Average gluteus medius RMS during stance phase (1-41% of gait cycle) and swing phase (60%-80% of gait cycle) can be seen in Figure 3.6. Stance phase gluteus medius RMS decreased with BWS (Table 3.1, $p < 0.001$). For swing phase gluteus medius RMS, there were no significant main effects.

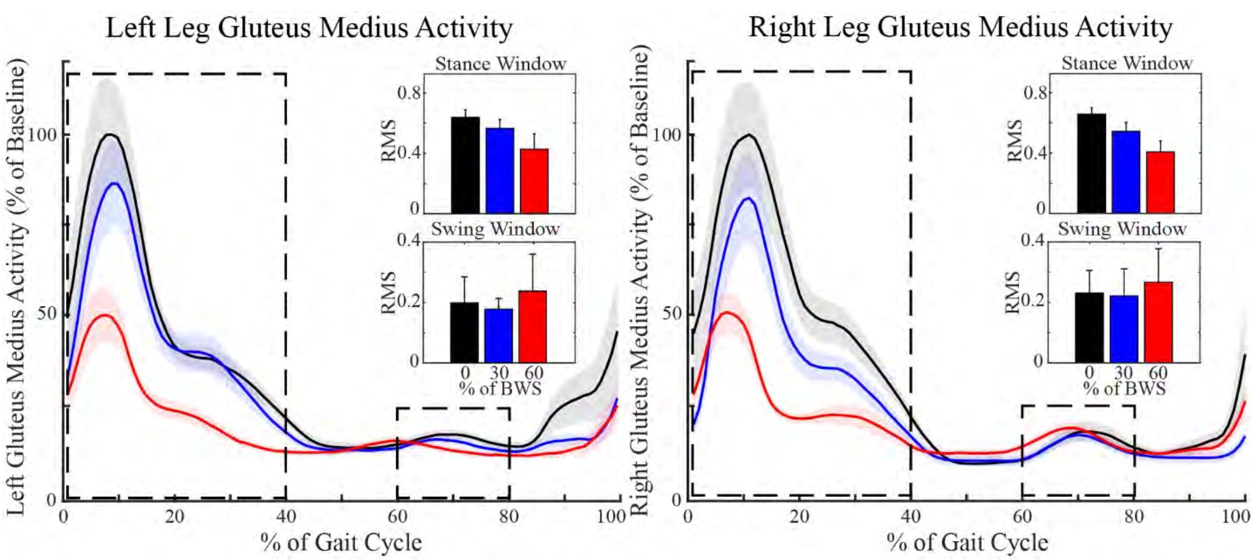

Figure 3.5 Effect of BWS on Gluteus Medius Activity Mean ± standard error of left and right leg gluteus medius activity for different levels of BWS (0% black, 30% blue, and 60% red). Dashed boxes represent time windows for statistical tests during the stance phase (1-40% of the gait cycle) and swing phase (60-80% of the gait cycle). Insert bar graphs shows RMS values ±

standard error for each level of BWS in the two time windows. * indicates there was a significant main effect of BWS (see Table 1 for further details).

Discussion

We found that BWS reduced frontal-plane COM excursion and velocity and induced COM trajectory asymmetries. Correlations between COM state and foot placement became weaker with BWS and asymmetric changes in foot placement position were observed. These findings strongly suggest that BWS changes how individuals' control frontal-plane COM walking dynamics.

The Effect of BWS on COM State

We observed significant changes in COM dynamics when walking with BWS. We found that as BWS increased, COM excursion and maximum COM velocity in the mediolateral and anteroposterior directions decreased. Previous studies have found similar reductions in COM excursion and acceleration [55], [58]. To our surprise, we also found that as BWS increased, the COM trajectory became increasingly asymmetric. Vertical-plane asymmetry (mediolateral vs anteroposterior position) was biased towards the right limb. The majority of our participants were right-limb dominant and may have favored their dominant side. However, this asymmetry may have also been induced by direct effects of the BWS system. BWS originating from a fixed overhead location will induce a pendulum effect that generates a lateral restoring force [1], [46], [74] to redirect the COM medially during gait. Although the magnitude of this restoring force is small (~1% of bodyweight at 60% BWS) [1], we attempted to minimize this effect by aligning the actuated trolley of our BWS system in the mediolateral direction to allow for lateral body movements. However, it is possible that delays or imperfections in the active tracking may have contributed to the asymmetries we observed.

The Effect of BWS on Stepping Strategy

We hypothesized that our participants would adopt a new stepping strategy when walking with BWS, where lateral foot placement would be less coordinated with lateral COM state. This hypothesis was supported. As the level of BWS increased, R^2 time-series correlations between foot placement and COM state weakened. The R^2 time-series correlations at 60% BWS are similar to those seen when individuals walk with external lateral stabilization [66]. If BWS produces a stabilizing effect it would reduce the demands for controlling frontal-plane dynamics. This effect might explain our finding that as BWS increased, there was a reduced need to coordinate foot placement relative to the COM state.

In addition, we found right-left asymmetries in foot placement similar to those observed in the COM trajectory. These asymmetries were associated with asymmetries in frontal-plane hip ROM, peak stance phase hip adduction angles, and peak swing phase hip abduction angles that increased as BWS increased. Overall, as BWS increased, the right leg was placed progressively more laterally to the COM, and the opposite occurred with the left leg. Despite these asymmetries, there were no differences in step width or step width variability between 0% and 60% BWS. These results conflict with our previous work that found wider, less variable steps as BWS increased [1], which may be due to differences in experimental setup.

Gluteus medius is the primary muscle for controlling frontal-plane dynamics and acts to redirect the COM during stance phase and the swing limb trajectory during swing phase. Recent studies have shown that increased activation of stance limb gluteus medius produces more medial foot placements and activation of swing limb gluteus medius produces more lateral foot [22], [104]. Similar to previous studies, we found that increased levels of BWS leads to reductions in stance limb gluteus medius activity [54], but increased levels of BWS did not change swing limb gluteus medius activity. We did not find any differences in gluteus medius activity between limbs, which suggests that the foot placement asymmetries may have been driven by the mechanical interaction between BWS and COM state. By decreasing the effects of gravity, BWS may discourage individuals from actively controlling COM state and foot placement position.

Conclusions

Although BWS is a valuable tool for early intervention gait training of several clinical populations [45], [71], [72], [92], it has not been shown to improve walking balance [43], [105]. The maintenance of lateral stability is a requirement for independent walking, and walking interventions should encourage individuals to adopt an appropriate stabilization strategy. When walking with BWS, healthy individuals adopted a compensatory stepping strategy that reduced the relationship between COM state and lateral foot placement. This finding suggests that BWS may encourage practice of stepping strategies that may have limited translation to control frontal-plane COM dynamics during real-world walking.

Acknowledgements

This work was supported in part by NIH training grant #T32EB009406-10. We would like to thank the members of the Human Agility Lab for their input and assistance with the study.

Chapter 4: Post-Stroke Adaptation of Lateral Foot Placement Coordination in Variable Environments

Abstract

Individuals with stroke often have difficulty modulating their lateral foot placement during gait, a primary strategy for maintaining lateral stability. Our purpose was to understand how individuals with and without stroke adapt their lateral foot placement when walking in an environment that alters center of mass (COM) dynamics and the mechanical requirement to maintain lateral stability. The treadmill walking environments included: 1) a Null Field – where no forces were applied, and 2) a Damping Field – where external forces opposed lateral COM velocity. To evaluate the response to the changes in environment, we quantified the correlation between lateral COM state and lateral foot placement (FP), as well as step width mean and variability. We hypothesized the Damping Field would produce a stabilizing effect and reduce both the COM-FP correlation strength and step width compared to the Null Field. We also hypothesized that individuals with stroke would have a significantly weaker COM-FP correlation than individuals without stroke. Surprisingly, we found no differences in COM-FP correlations between the Damping and Null Fields. We also found that compared to individuals without stroke in the Null Field, individuals with stroke had weaker COM-FP correlations (Paretic < Control: $p = 0.001$, Non-Paretic < Control: $p = 0.007$) and wider step widths ($p = 0.001$). Our results suggest that there is a post-stroke shift towards a non-specific lateral stabilization strategy that relies on wide steps that are less correlated to COM dynamics than in individuals without stroke.

Introduction

People with an intact neuro-musculoskeletal system exhibit step-to-step changes to mediolateral foot placement in response to small changes to their center of mass (COM) state (position and velocity) during walking [17]. For example, if the COM is positioned more laterally during the swing phase, people often respond with a wider step. This step-to-step coordination aids in the maintenance of frontal plane stability. Specifically, coordinated modulation of lateral foot placement will scale the medially-directed gravitational moment created about the ankle joint to the immediate demands imposed by the COM state [11]. During the single-limb support phase, active adjustment of lateral foot placement combined with the passive dynamics of the swing limb leads to high correlations between lateral COM state and impending lateral foot placement. By mid-stance, over 80% of the variance in step-to-step lateral foot placement can be explained by lateral COM state, which approaches 100% by initial contact as dictated by passive dynamics [12], [17], [66], [98]. Collectively, this research suggests that coordinated changes in lateral foot placement resulting from an interplay of both active control strategies and passive dynamics may be an important mechanism for maintaining mediolateral gait stability.

However, for individuals recovering from stroke, the capacity to make coordinated adjustments to lateral foot placement in response to changes in COM state may be impaired. Indeed, recent work by Stimpson et al. found evidence that within individuals with stroke, this correlation is weaker for paretic steps than for non-paretic steps [33]. For this population, challenges in coordinating lateral foot placement may be a result of both sensory-motor deficits and changes in lower-body mechanical properties (e.g., joint stiffness). Individuals with stroke exhibit deficits in lateral foot placement control [38], greater variability when prescribed a particular step width

[37], and proprioception deficits in muscles that act in the frontal plane [39], each of which may impair the ability to actively coordinate lateral foot placement and COM state. Neurologic changes following a stroke, such as abnormal torque synergies [106], [107] and spasticity [108], alter the coordination of the paretic limb. These neurologic changes affect the available range of motion, force generation, and an individual's ability to independently control degrees-of-freedom in the lower extremity of both the paretic and non-paretic extremities. In addition, because the lower extremity is mechanically linked to the pelvis, passive transfer of energy across body segments (e.g. from the pelvis to the thigh segment) can also contribute substantially to the coordination between lateral foot placement and COM state [12], [17], [98]. As such, neurologic changes following a stroke may impose limits on this transfer of energy between segments. Together, these factors suggest that the coordinated variability in lateral foot placement in response to fluctuations in COM state will be impaired in post-stroke populations compared to their counterparts without stroke.

The level of step-to-step coordination between lateral COM state and lateral foot placement responds to the continuous demands to maintain lateral stability. Specifically, when the demands to maintain lateral stability are reduced by external lateral forces that resist lateral COM excursions [34], [36] or velocity [62], individuals exhibit decreases in step width and step width variability [34], [36], [62], [109], as well as a weaker correlation between lateral COM state and lateral foot placement [66]. Conversely, when individuals walk in environments that increase the demands to maintain lateral stability (e.g. a movement amplification force field that amplifies lateral COM velocity), individuals exhibit increases in step width and step width variability [61], [63], as well as a stronger correlation between lateral COM state and lateral foot placement [63].

It is unknown if the correlation between lateral COM state and lateral foot placement will also adapt to the altered demands to maintain lateral stability in individuals with stroke. A recent study found that some individuals with stroke reduced their step width when walking with external lateral stabilization, while others either increased or did not change their step width [64]. The post-stroke coordination of lateral COM state and lateral foot placement has not been examined while walking with external stabilization, but the inconsistent findings in step width changes suggest that the impaired neuromuscular system is limited in its ability to adapt foot placement in response to significant changes in lateral stability demands.

Therefore, our purpose was to better understand how adaptable lateral foot placement is to changes in COM state for individuals with and without a stroke. To accomplish this, individuals with and without stroke walked at their preferred speed during normal steady-state walking (Null Field) and in a velocity-based Damping Field that resists lateral COM velocity and reduces the demands to maintain lateral stabilization. It has been recently shown in neurologically-intact individuals that external fields which limit COM excursion weaken the correlation between COM state and lateral foot placement [66]. Although a velocity-based field is mechanically different from a position-based field in the application of lateral forces, past research suggests that lateral foot placement variability will be affected by manipulation of COM position and velocity [17], [63], [66]. In an environment that reduces the stabilization requirements, fewer or less precise adjustments in compensatory lateral foot placement may be necessary. Therefore, we hypothesized that compared to the Null Field, the correlation between lateral COM state and lateral foot placement would be weaker in the Damping Field across both populations. In addition, we expected the Damping Field to lead to reductions in step width and step width variability compared to the Null Field. We also hypothesized that individuals with stroke would exhibit a weaker correlation between lateral COM state and lateral foot placement across fields compared to their peers without stroke.

Methods

Participants

We collected data and analyzed data from a total of 18 participants. The participants included nine individuals with chronic stroke (7 male, 59 +/- 7 years old) and nine individuals without stroke (9 male, 61 +/- 6 years old). Both Northwestern University and Edward Hines Jr. Veterans Administration Hospital Institutional Review Boards approved the study protocol and procedures. All participants provided written, informed consent prior to data collection.

Data for individuals without stroke were collected as a part of a separate study [65] that, as described below, used a similar experimental protocol but included small differences in inclusion/exclusion criteria and data collection methods. Individuals with stroke were included if they experienced a stroke at least one year prior to data collection, had unilateral paresis, were not using medications that affect walking ability or balance, and had the ability to walk continuously for 6 minutes without the use of an assistive device. Individuals without stroke were included if they were able to walk 10 minutes without undue fatigue, were not using medications that affect walking ability or balance, and did not have any musculoskeletal or neurological pathology that would affect their walking ability or balance.

General demographic information (all participants) and clinical outcome measures characterizing walking speed (Self-Selected and Fast 10-Meter Walk Test [110]), postural balance (Berg Balance Scale [111]), and lower limb function (Lower Extremity Fugl-Meyer Assessment [112]) for participants with stroke are included in Table 1.

Table 4.1 Participant Demographics and Clinical Outcome Measures

	Sex	*Age*	*Paretic Side*	*Preferred Speed (m/s)*	*10MWT Self*	*10MWT Fast*	*BBS*	*FMA LE*
Stroke								
S1	M	61	R	0.58	0.38	0.68	55	23
S2	M	58	R	0.40	0.40	0.53	53	17
S3	M	47	L	0.49	0.44	0.67	55	23
S4	F	64	R	0.18	0.31	0.37	50	19
S5	F	68	R	0.36	0.38	0.51	55	19
S6	M	49	R	0.72	0.53	0.76	45	21
S7	M	63	R	0.22	0.22	0.14	32	19
S8	M	64	L	0.36	0.69	0.86	53	23
S9	M	58	L	0.58	0.69	0.93	51	22
Average		59		0.43	0.45	0.60	50	21
STD		7		0.17	0.15	0.23	7	2
Control								
C1	M	64		1.07	1.30	1.70		
C2	M	71		1.03	1.23	2.03		
C3	M	62		0.94	1.30	1.70		
C4	M	56		1.12	1.60	2.13		
C5	M	67		0.67	1.13	1.81		
C6	M	64		0.85	1.10	1.50		
C7	M	56		1.30	1.50	2.20		
C8	M	55		1.21	1.40	1.90		
C9	M	50		1.21	1.50	2.20		
Average		61		1.04	1.34	1.91		
STD		6		0.19	0.16	0.24		

Experimental Setup

Participants performed a series of walking trials on an oversized treadmill (2.6 m long x 1.4 m wide) (Tuff Tread, Willis, TX). Participants wore a trunk harness attached to a passive overhead support system that did not provide bodyweight support (Aretech, Ashburn, VA).

During select walking trials, a Damping Field applied external lateral forces to the pelvis. The Damping Field was produced using a cable-driven robotic device, the Agility Trainer, which uses the excursion of the linear motors to estimate COM state [61]. Participants walking in the Damping Field experienced a continuous, lateral force proportional in magnitude and opposite in direction of their real-time lateral COM velocity. The gain for the Damping Field was 50 Ns/m, which was selected as previous research has found that this gain was sufficient to produce significant changes in lateral foot placement [109].

To measure gait kinematics, we used a 12-camera motion capture system (Qualisys, Gothenburg, Sweden) to record the 3D positions of 33 retro-reflective markers at 100 Hz. Markers were affixed bilaterally to the anterior and posterior iliac spines, greater trochanters, lateral epicondyle of the knee, lateral malleoli, calcanei, and the 2nd and 5th metatarsals. Additional 4-marker tracking clusters were affixed bilaterally to the participant's thigh and shank, and a single marker was placed on the sternum.

Protocol

Prior to walking trials, a licensed physical therapist performed clinical outcome measures to assess walking and balance function for participants with stroke. These clinical assessments included the Lower Extremity Fugl-Meyer Assessment (LE-FMA) [112], Berg Balance Scale (BBS) [111], and the 10 Meter Walk Test (10MWT) [110] at both self-selected and fast speeds. Clinical outcome measures for individuals without stroke included only the 10MWT.

Next, participants with stroke performed walking trials to determine their preferred treadmill walking speed. Participants were asked to subjectively compare gradual increases and decreases in speed until they reached a steady-state speed they felt most comfortable walking at on the treadmill.

Finally, each participant with stroke performed a randomized series of walking trials. Individuals walked at their preferred walking speed with two force field conditions: 1) Null – no applied forces, and 2) Damping – the velocity-based damping field described above was applied. Participants with stroke repeated each condition twice. During each trial, participants walked for a total of 5 minutes: 3 minutes to familiarize and remove any learning effects from previous trials, immediately followed by 2 minutes of walking, during which data were collected. Participants did not use handrails or assistive devices during the walking trials; however, if participants normally wore an ankle-foot orthosis for community ambulation, they were allowed to wear it during the trials (n=5).

Individuals without stroke performed treadmill walking trials at their preferred speed in Null and Damping Fields using similar protocols to those described above for individuals with stroke. Individuals without stroke walked for 200 continuous steps each trial and performed additional treadmill walking trials that were not analyzed for the current study [65].

Data Analysis

We analyzed walking data collected from individuals with and without stroke. To make comparisons between trials and groups, we chose to analyze only the last 50 steps from each trial. This ensured that any measures that were sensitive to the number of steps analyzed were consistent across all participants.

Kinematic data were processed with Qualisys Track Manager (Qualisys, Gothenburg, Sweden), Visual3D (C-Motion, Germantown, MD), and custom MATLAB (Mathworks, Natick, MA) scripts. Marker data were low-pass filtered (4th-order Butterworth, 6 Hz cut-off frequency) and gap-filled in Visual3D. Gait events (foot-off and initial contact) were identified based on the vertical position of the markers on the calcaneus and 5th metatarsal. Timing of gait events was visually checked for accuracy. Additionally, COM position and velocity were calculated in Visual3D using a pelvis segment that was defined by bilateral markers on the iliac crests and greater trochanters. This estimate of COM state has been shown to minimize bias in estimating COM dynamics [101].

To investigate stepping strategy, a regression equation was used to predict the next mediolateral foot placement based on the COM state (position, relative to the stance foot, and velocity) at discrete time points during the preceding swing phase [17]. This method, described by Wang & Srinivasan, calculates the coefficient of determination (R^2) as the ratio between the variance in the dependent variable (lateral foot placement position) that is predicted by the independent variables (lateral COM position and velocity) [17]. As the dimensionless unit R^2 increases, it indicates that there is a smaller difference between the predicted and actual foot placement. Therefore, a greater R^2 indicates a stronger correlation between COM state and foot placement. To model this stepping strategy, we adapted the following regression equation:

$$FP_x = \beta_1(i) \cdot COM_x(i) + \beta_2(i) \cdot COM_v(i) + \varepsilon(i)$$

where FP_x is lateral foot placement position relative to the contralateral stance foot, COM_x is the lateral COM position, COM_v is the lateral COM velocity, β_1 and β_2 are regression coefficients, ε is the error, and i is a discrete time point during the preceding swing phase. The independent variables (COM state) were demeaned prior to their use in the regression equation. This regression was repeated for every 2% of the preceding swing phase. The resulting time-series quantification of the R^2 values between the predicted and actual lateral foot placement positions was used as the primary outcome measure. For participants with stroke, separate R^2 time-series were computed for the paretic and non-paretic extremities, and a single R^2 time-series was computed for the control extremities. For nomenclature, a paretic R^2 time-series refers to steps where the paretic extremity is in the swing phase.

We also calculated two additional gait metrics that could provide insight into the strategies used by participants to maintain frontal plane stability. These gait metrics included step width mean and variability. Step width was calculated as the mediolateral distance between the calcaneus markers at initial contact. For individuals with stroke, this metric was separated into a paretic and non-paretic step width (i.e., paretic step width was the lateral distance from the calcaneus of the paretic extremity at initial contact to the position of the non-paretic calcaneus at the preceding initial contact). Step width variability was calculated as the standard deviation of step width.

Statistical Analysis

To compare stepping strategies, we used MATLAB to perform a two-way ANOVA using a statistical parametric mapping (SPM) approach [66]. For a given dataset, the SPM approach regards the data as a vector field which changes in time or space and uses principles from random field theory to calculate the probability that changes in the vector field are due to chance fluctuations [103]. This approach can compare features over the entire time-series and has been shown to be generalizable to a variety of 1-, 2-, and 3-dimensional biomechanical datasets [113], [114]. We performed a two-way ANOVA with fixed effects of limb (Paretic, Non-Paretic, and Control), field (Null and Damping), and the interaction between limb and field. The output of SPM produces a F-value for each sample of the R^2 time-series. If the F-value for a given time point crosses a set threshold value, corresponding to $\alpha = 0.05$, it indicates a statistically significant effect. If there was a significant main effect of limb or an interaction of limb and field, Bonferroni-corrected pairwise comparisons were made to determine significant differences.

To compare gait metrics, we used SPSS (IBM, Armonk, NY) to create separate linear mixed effects models for the two variables: step width mean and variability. Fixed effects included limb (Paretic, Non-Paretic, and Control), field (Null and Damping), and the interaction between limb and field. Random intercepts allowed each participant to deviate from the main intercept. For participants without stroke, a paired t-test was performed to determine if there was a significant effect from limb side; otherwise, data for the group were collapsed across limbs. If there was a significant interaction between limb and field, Bonferroni-corrected pairwise comparisons were made to determine the significant pair(s) of fields (Null vs. Damping) within each limb and of limbs (Paretic vs. Non-Paretic, Paretic vs. Control, and/or Non-Paretic vs. Control) within each field. Significance was set to $p<0.05$ for all tests.

Results

Participant Demographics

The nine participants with stroke were 59 ± 7 years old and 7 males/2 females with a preferred treadmill walking speed of 0.43 ± 0.17 m/s. Gait speeds (as indicated by 10-Meter Walk Test) suggest that participants with stroke would be classified as limited community ambulators (gait speed 0.4-0.8 m/s [115]). Scores from the Berg Balance Scale indicate that participants were not at increased risk for falling (Berg > 44 [116]). Scores from Lower Extremity Fugl-Meyer suggest that participants were moderately impaired (LE-FM between 19 and 27, [117]). The nine participants without stroke were 61 ± 6 years old and all males with a preferred treadmill walking speed of 1.04 ± 0.19 m/s. Participant demographics and clinical outcome measures for both groups are included in Table 4.1.

Damping Field

Lateral COM velocity throughout the swing phase was reduced in the Damping Field in comparison to the Null Field (Figure 4.1). The differences in velocity between fields were greatest at foot-off and initial contact. For the participants without stroke, lateral COM velocity decreased by 24% at foot-off and 46% at initial contact in the Damping Field compared to the Null Field. Similar reductions were seen in the participants with stroke during the swing phase of the non-paretic extremity, where COM velocity decreased by 41% at foot-off and 54% at initial contact in the Damping Field compared to the Null Field. During the swing phase of the paretic extremity, lateral COM velocity decreased by 24% for both foot-off and initial contact in the Damping Field compared to the Null Field.

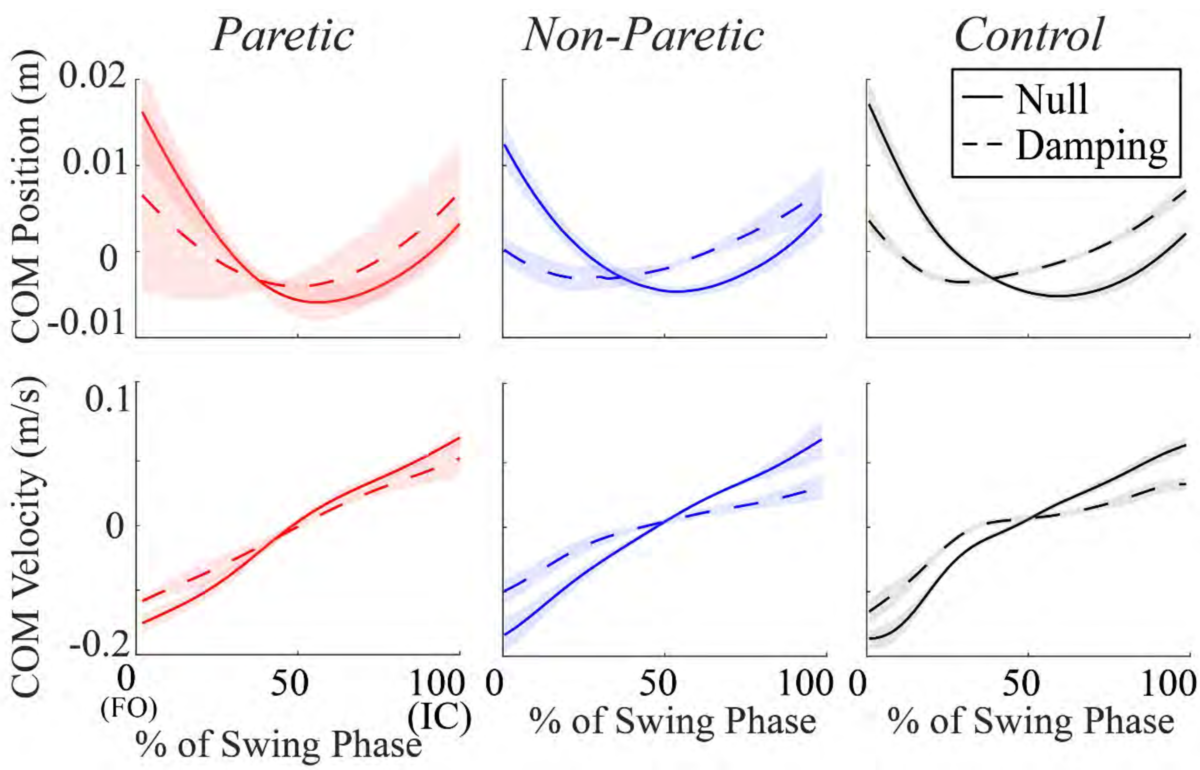

Figure 4.1 The Effect of the Damping Field on COM Position and Velocity Mean ± standard error of lateral COM position and velocity during swing phase for each limb (Paretic, Non-Paretic, and Control) and field (Null: solid, Damping: dashed)

The Damping Field also affected lateral COM position, primarily by creating a change in the relative time when maximum COM excursion occurred during the swing phase. Across all groups, maximum lateral COM excursion during the swing phase was earlier in the Damping Field than in the Null Field. For the participants without stroke, peak COM excursion occurred at 60% of the swing phase in the Null Field but at 30% of the swing phase in the Damping Field. A similar shift occurred during the swing phase of the non-paretic extremity, shifting the time-point of peak COM excursion from 54% of the swing phase in the Null Field to 34% in the Damping Field. A smaller shift occurred during the swing phase of the paretic extremity, where the time-point of peak COM excursion shifted from 56% of the swing phase in the Null Field to 50% in the Damping Field.

Table 4.2 R^2 Data

	Null Field		**Damping Field**	
	Foot-off	*Initial Contact*	*Foot-off*	*Initial Contact*
Paretic	0.35 ± 0.17	0.44 ± 0.25	0.47 ± 0.21	0.54 ± 0.25
Non-Paretic	0.40 ± 0.18	0.50 ± 0.18	0.53 ± 0.21	0.54 ± 0.19
Control	0.46 ± 0.11	0.76 ± 0.12	0.55 ± 0.15	0.74 ± 0.04

Stepping Strategy

To determine how participants adapted their stepping strategy between limbs and within the fields, we used a regression equation to predict lateral foot placement position based on the lateral COM state (position and velocity) during the preceding swing phase of gait (Figure 4.2). This regression equation produced a R^2 time-series that represents the amount of the variance in foot placement that can be explained by COM state. For all limbs, the R^2 time-series values tended to increase throughout the preceding swing phase of gait. R^2 values at the beginning and end of the swing phase are included in Table 4.2.

There was a significant main effect of limb ($F(1,17) = 4.451$, $p = 0.01$) between 30%-100% of the swing phase, but not of field or the interaction between the two (Figure 4.3). Comparing the R^2 time-series values between the limbs (Figure 4.4), there were significant differences between the paretic and control limbs (Control > Paretic, $t(1,29) = 3.038$, $p = 0.001$) from 28%-100% of the swing phase and between the non-paretic and control limbs (Control > Non-Paretic, $t(1,29) = 3.038$, $p = 0.007$) from 62%-100% of the swing phase. There were no significant differences in R^2 time-series between the paretic and non-paretic limbs.

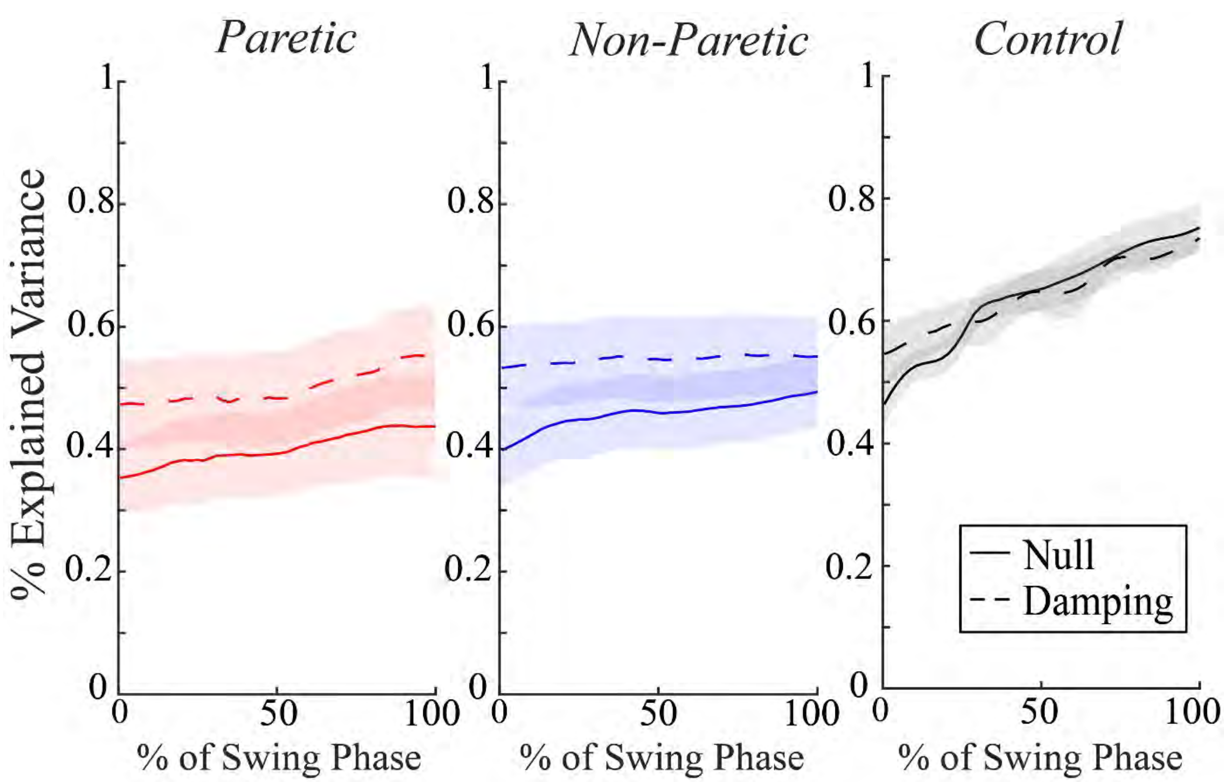

Figure 4.2 Correlation of Lateral COM State to Lateral Foot Placement Position Mean ± standard error (shaded region) of R^2 time-series showing capacity of COM state (position and velocity) to predict subsequent lateral foot placement for limbs (Paretic, Non-Paertic, and Control) and fields (Null: solid and Damping: dashed) throughout the swing phase of gait.

Gait Metrics

To determine how participants adapted their gait between limbs (Paretic, Non-Paretic, and Control) and within the fields (Null and Damping), we examined the average step width and step width variability (Figure 4.5). For step width, there was a significant effect of limb (p=0.001), but not of field (p=0.958) or the interaction of limb and field (p=0.696). Pairwise comparisons found wider steps for the paretic and non-paretic extremities compared to the control extremity (Paretic > Control, p=0.001; Non-Paretic > Control, p=0.001). For step width variability, there were no significant effects of limb (p=0.283), field (p=0.566), or the interaction of limb and field (p=0.068).

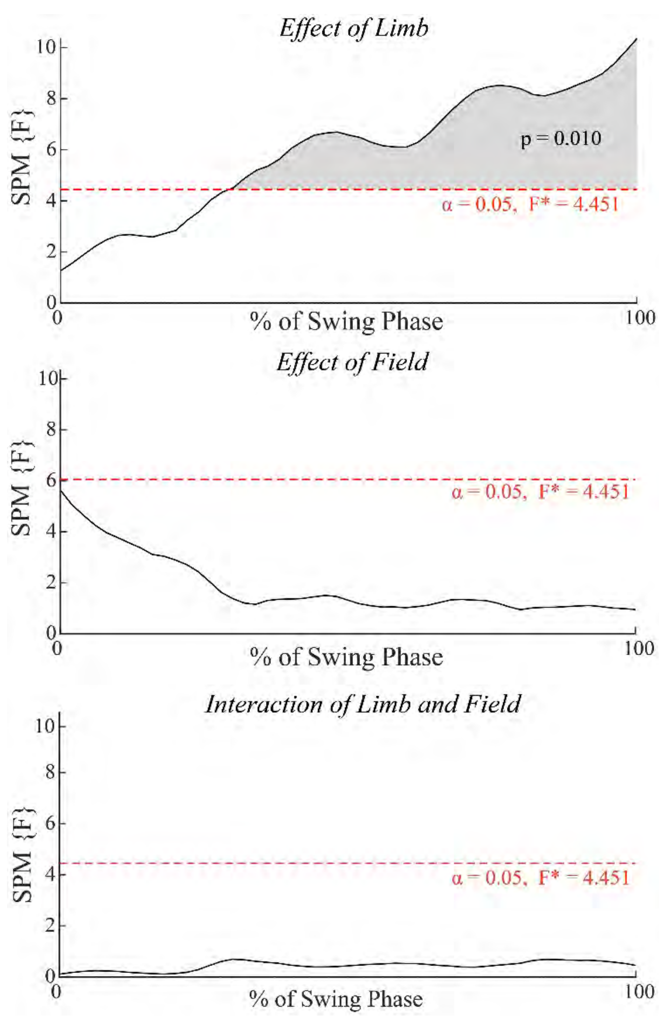

Figure 4.3 2-Way ANOVA of Limb and Field 2-way ANOVA with main effects of limb (Paretic, Non-Paretic, Control) and field (Null and Damping) and the interaction of limb and field on R^2 time-series. The dashed red line indicates the threshold F-value, corresponding to an $\alpha = 0.05$, where values above this threshold are statistically significant. Shaded regions indicate significant effects for the corresponding portion of the swing phase.

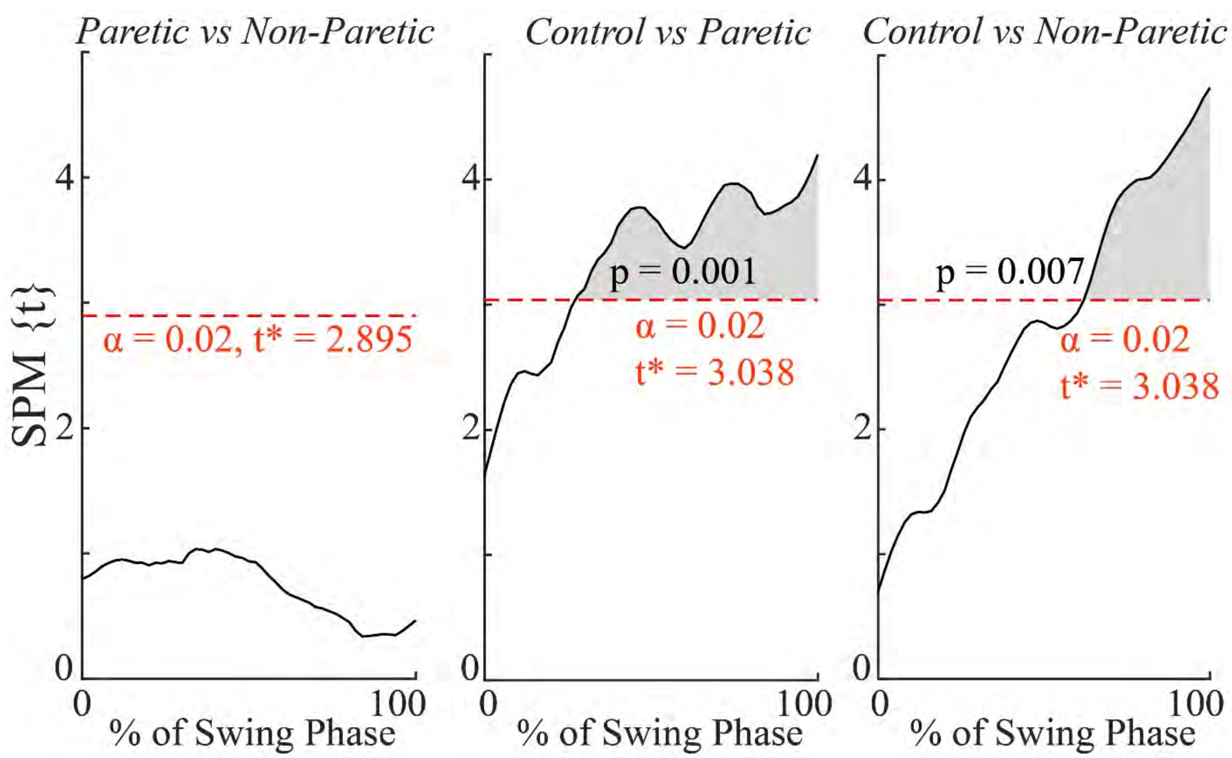

Figure 4.4 Post-hoc Comparisons of Within Limb Effects Differences in R^2 time-series between limb pairs (Paretic vs. Non-Paretic, Control vs. Paretic, Control vs. Non-Paretic) with Bonferroni corrected p-values. The dashed red line indicates the threshold t-value, corresponding to an $\alpha = 0.02$, where values above this threshold are statistically significant. Shaded regions indicate significant effects for the corresponding portion of the swing phase.

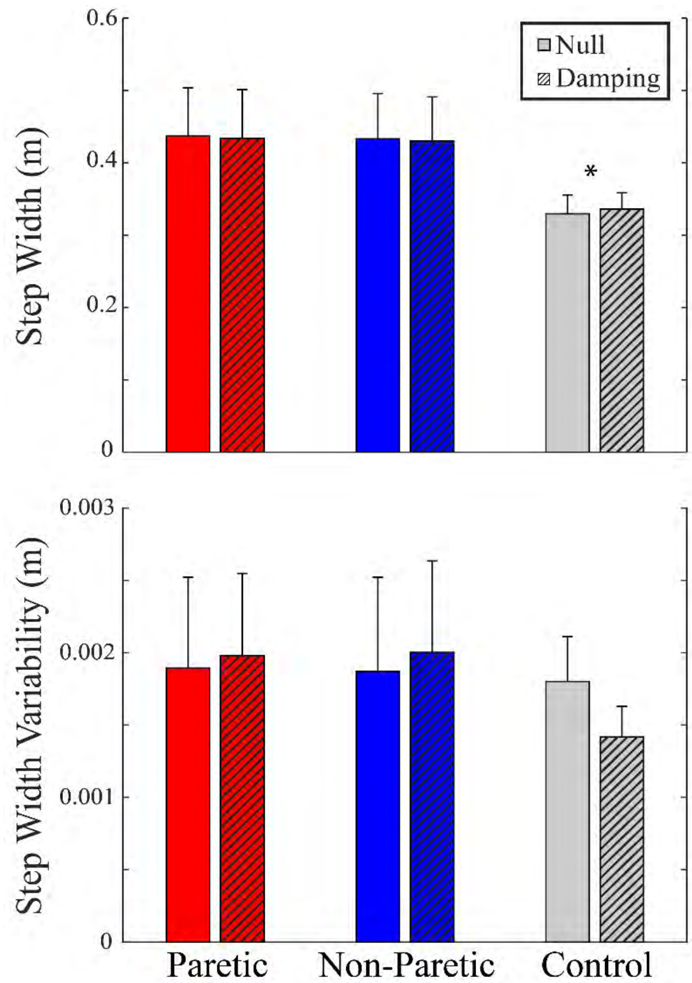

Figure 4.5 Effect of Damping Field on Step Width and Step Width Variability Gait metrics means ± standard errors of step width and step width variability. * indicates the Control limb exhibited a significantly smaller step width than the Paretic or Non-Paretic limbs (Paretic > Control, p = 0.001; Non-Paretic > Control, p = 0.001)

Discussion

The purpose of this study was to investigate if individuals with stroke adapt their step-to-step coordination between lateral COM state and lateral foot placement in response to significant changes in lateral stability demands. Here we imposed an external force field that resisted lateral COM velocity to probe the effect of COM state changes during walking on lateral foot placement. We investigated if individuals with and without stroke adapted their stepping strategy (correlation of lateral COM state and lateral foot placement position) and gait parameters (step width mean and variability) in response to a Damping Field. We hypothesized that in comparison to the Null field, the Damping Field would result in a weaker correlation between COM state and foot placement, as well as a reduction in step width mean and variability. This hypothesis was not supported, suggesting that there is a complex relationship between an individual's stepping strategy and the demands to maintain lateral stability. Additionally, we hypothesized that the correlation between COM state and foot placement would be weaker for individuals with stroke (both paretic and non-paretic extremities) than for individuals without stroke. This second hypothesis was supported and highlights the differences in mechanisms people with and without stroke use to maintain gait stability.

Differences between Null and Damping Fields

Our hypothesis that correlations between lateral COM state and lateral foot placement would be weaker in the velocity-resistant Damping Field than in a Null Field was not supported. This hypothesis was based on previous research suggesting that the strength of the coupling between lateral COM state and lateral foot placement scale to the requirements to maintain lateral stability. We quantified the effect of the Damping Field on the relationship between lateral COM state and lateral foot placement by comparing a full swing-phase R^2 time-series between fields but found no significant effect of field. For the paretic and non-paretic limbs, R^2 values were greater at foot-off and initial contact in the Damping Field than in the Null Field and consistently higher throughout the swing phase (Fig. 2). These results suggest a complex relationship between an individual's stepping strategy and the demands to maintain lateral stability.

Some previous experimental and simulation studies have suggested that there is an active control strategy for maintaining lateral stability that adjusts lateral foot placement relative to the COM state. By recruiting frontal plane abductors in the stance [104] and swing limbs [22], [104], an individual can alter their step-to-step lateral foot placement to account for lateral COM state. This strategy is utilized during the early to mid-swing phase when the nervous system can make corrections to lateral foot placement, whereas the latter half of the swing phase is primarily influenced by the passive dynamics of the swing limb. Previous studies in individuals without neurological injury have found that this correlation is adaptable to the demands to maintain lateral stability through the application of external lateral forces, where external lateral stabilization decreases the correlation [66] and lateral movement amplification increases the correlation [63], [65], although these correlation changes may be driven by manipulation of COM dynamics by the forces themselves. We expected the Damping Field to disrupt the coordination between the lateral COM and lateral foot placement; however, the Damping Field did not lead to differences in correlations or step width between the two fields. We suspect that the lack of changes in the Damping Field may be partly explained by a decreased sensitivity to manipulations of lateral COM velocity compared to COM position. To determine the relative sensitivity of the two regression coefficients, we examined the regression coefficients between the two fields (and the three limbs) and found that they were quite similar. We found that β_1 values (COM position) were consistently positive ($\beta_1 \approx 0.9$ at mid-stance) and significant throughout the swing phase for all conditions and limbs. β_2 values (COM velocity) had smaller magnitudes ($\beta_2 \approx 0.1$ at mid-stance) and were non-significant during the swing phase from 20-60% depending upon the condition and limb. The magnitudes of our regression coefficients are smaller than has previously been reported at mid-stance ($\beta_1 \approx 2.01$ and $\beta_2 \approx 0.44$) [17], meaning deviations of the COM position or velocity led to smaller changes in lateral foot placement position than previous reports. For both individuals with and without stroke, COM position explains most of the variability in lateral foot placement position. Our participants were older than those in previous studies and may have experienced age-related changes in COM stability [118] and hip abductor strength [119] that affected the relationship between lateral COM state and lateral foot placement position. Ultimately, this reduced sensitivity to COM velocity deviations may have contributed to the lack of changes seen in the correlations when walking in our velocity-based Damping Field compared to previously-examined position-based fields.

Differences between Limbs

In support of our second hypothesis, we found individuals with stroke had significantly weaker correlations between lateral COM state and lateral foot placement compared to individuals without stroke across the fields. These differences between the limbs emerged at ~28 and 62% of the swing cycle. In addition, while both groups increased R^2 values from foot-off to initial contact, the magnitude of this increase was 2-3x greater in the individuals without stroke. These results suggest that the strategies used to coordinate COM state and foot placement are different for individuals with and without stroke. We suspect the minimal changes in correlations from the start to the end of swing phase for the paretic and non-paretic extremities is indicative of a compensatory, non-specific stepping strategy.

If an individual is using a specific strategy to coordinate foot placement, one would expect the correlation between lateral COM state and lateral foot placement to strengthen throughout the swing phase, particularly during the first half. For an individual with an intact neuro-musculoskeletal system, there would be sufficient time to accurately analyze their COM state and make corrections to their swing-limb trajectory with an appropriate foot placement. Our observations of individuals without stroke found that R^2 values in the Null Field increased from 0.46 at foot-off to 0.76 at initial contact, suggesting that this group may be using this specific control strategy.

If an individual were relying on a non-specific strategy for maintaining lateral stability by selecting an indiscriminately wide step width, COM state would not become increasingly predictive of foot placement during the progression through the swing phase. This strategy may decrease the contribution of passive dynamics on the observed correlations. Other recent work has also suggested similar increases in reliance on non-specific strategies to maintain gait stability in individuals with incomplete spinal cord injury [62], [109] and stroke [33]. Our observations of individuals with stroke were consistent with this non-specific strategy, as participants in the current study took significantly wider steps than their peers without stroke and their R^2 values only increased by 0.09 or 0.10 from foot-off to initial contact for the paretic and non-paretic extremities respectively when walking in the Null Field. A reliance on non-specific strategies may be beneficial to individuals with stroke, who may have sensory [39] and motor deficits [38], [69] that impair their ability to control lateral foot placement during walking. These deficits include a reduced ability to target and choose an appropriate foot placement [37],

[69][37], [69][37], [69][37], [69] ability to regulate prescribed step widths [38][38][38][38], and proprioception of lateral hip musculature [39]. They are further limited by neurologic deficits, such as abnormal lower extremity torque synergies [106], [107], increases in spasticity [108], the adoption of compensatory paretic hip circumduction and non-paretic hip hiking [120], all of which may limit the coordination of lower extremity trajectories and torques. Our results suggest that the coordination of lateral foot placement relative to lateral COM state is significantly different between individuals with and without stroke. Our results differ from previous work that found individuals with stroke behave more similarly to those without stroke [33], although these differences may be due to the heterogeneity of the post-stroke population that possess a wide range of functional abilities.

Limitations

There are two important limitations to consider when interpreting these results. The first is the small sample size (n = 9 in both groups). Depending on the variable of interest, we found statistically significant differences between groups with significance that ranged from p = 0.05 to 0.001; however, these results should be interpreted with caution. Our small sample size limits the practical significance of our results, and future work that investigates the effects of the Damping Field with a larger sample size should be conducted to better distinguish the differences in stepping strategy between individuals with and without stroke.

The second limitation is the differences in average gait speeds between the two groups. Individuals without stroke walked with average gait speeds that were twice as fast as individuals with stroke. Previous research has found that correlations between lateral COM state and lateral foot placement are sensitive to gait speed in individuals with an intact neuro-musculoskeletal system [65], [99]. However, while the group differences in R^2 in early swing phase may be greatly influenced by speed, the differences in late swing phase may be more significantly affected by differences in stepping strategy and neurologic changes following a stroke. Here we were interested in measuring the differences in stepping strategy at an individual's preferred walking speed, but further work should investigate how these correlations are influenced by changes in gait speed for individuals with stroke.

Clinical Implications

Individuals with stroke often have difficulty maintaining gait stability, putting them at a higher risk for falling than their peers without neurological injury [7]. Challenges to stability post-stroke can arise from internal factors, such as distal lower extremity weakness [121], spasticity [108], and abnormal torque synergies [106], [107]. These internal factors limit the adaptability of the post-stroke neuromuscular control system to meet the mechanical requirements of gait and contribute to the observed adoption of abnormal gait patterns [122] and compensatory behaviors [120]. It is likely that these internal factors and their requisite compensations may impede the capacity of post-stroke gait to respond to changes in the external environment using stabilization strategies that require coordination between the COM and lower extremities. Our results suggest that in comparison to their peers without stroke, individuals with stroke utilize a more non-specific strategy for maintaining lateral stability. Rather than making step-to-step adjustments of lateral foot placement that are coupled to their ongoing COM dynamics, individuals with stroke demonstrate a compensatory strategy that prioritizes a wider step width. While this strategy may provide a robust solution for maintaining lateral stability, it has trade-offs, including increasing the metabolic energy cost of walking [123] and limiting maneuverability [25]. Interventions that can strengthen the correlation between COM state and lateral foot placement may be beneficial for practicing some desirable locomotor stability strategies. For example, the application of force fields at the COM [65] or at the swing limb [42] can effectively influence the relationship between the COM dynamics and foot placement. For clinical populations that utilize compensatory, non-specific strategies for maintaining lateral stability, these interventions may have the potential to be used to improve coordinated stepping behaviors.

Chapter 5: Conclusion

Summary of Work

This dissertation aims to characterize how gait training devices interact with the control of stability. I also provide a framework for developing a new, stability-based gait training intervention for people post-stroke.

In my first aim, I investigated how a BWS system affects the control of gait stability. I found that BWS created a dynamic environment which changes the locomotor control requirements needed to maintain stability. In response to BWS, I observed that young adults made complex changes in their walking patterns. Specifically, when walking with BWS participants adapted a gait pattern with wide steps and low step width variability, which are generally associated with unstable and stable gait patterns, respectively. The gait patterns practiced with BWS may be poorly suited for the stability requirements of real-world walking. Clinicians who use BWS for gait training should consider that BWS may encourage a practice of gait patterns that minimize active control of lateral stability.

In my second aim, I further characterized the effects of BWS on the dynamics of walking and determined how BWS affects stepping coordination. As hypothesized, high levels of BWS reduced coordination between the COM state and foot placement. I also found that as the level of BWS increased the COM trajectory became more asymmetric in the mediolateral direction. I found a similar asymmetry in foot placement position, where the right leg took increasingly more lateral steps than the left leg. These results suggest that BWS may encourage a compensatory and non-specific stepping strategy to control of locomotor stability. The conclusion of this study provides further evidence that the gait patterns practiced with BWS may be poorly suited for the stability requirements of real-world walking.

Together, my first two aims demonstrate that BWS substantially changes frontal plane stepping behavior. Maintaining stability is a requirement for independent walking, and gait training interventions should encourage patients to adopt an appropriate stepping strategy. The non-specific stepping strategy exhibited when walking with BWS may have limited translation for controlling the dynamics of real-world walking.

In my third aim, I demonstrate a new type of gait training device for people post-stroke that directly challenges the control of stability by the locomotor system. I found that stepping strategies were persevered when participants walk with lateral assistance. The Damping Field was able to successfully change the dynamics of walking and tended to improve stepping coordination for participants with stroke. This work also identified that when compared to non-impaired individuals that individuals with stroke use a non-specific strategy to maintain stability that prioritizes a wide step width with reduced coordination between the COM and swing foot. Gait training with progressive lateral assistance may be a useful method for improving coordination for individuals with stroke and maybe applicable to other clinical populations that use non-specific stepping strategies, such as individuals with spinal cord injury. Additionally, the increased coordination may improve translation to real-world walking.

In summary, this work describes a method for evaluating how gait training interventions interact with the stepping strategies used to control stability. Maintaining stability is a basic requirement for independent ambulation, and it is crucial to understand if gait training interventions encourage stepping strategies that are appropriately suited for the dynamics of real-world walking. For individuals with stroke who use non-specific stepping strategies, progressive lateral assistance can create dynamic environments that encourage more coordinated stepping and may produce more active control of stability.

I propose that moving forward, gait training interventions should systematically challenge the control of stability and allow for practice of more coordinated stepping strategies. This dissertation demonstrates that walking with lateral assistive forces may be able to improve coordination in people post-stroke and unlike BWS does not disrupt the typical coordination between lateral foot placement and COM state observed during real world walking. Future work should determine if gait training with progressive lateral assistance produces beneficial and lasting effects for people post-stroke during community ambulation.

Future Directions

The findings of this dissertation and the proposed stability-based framework for gait training can be extended in several clinically relevant directions.

The first two aims of this dissertation demonstrate how BWS interacts with the control of gait stability and stepping coordination for healthy adults. A natural extension of this work is to determine how BWS affects frontal plane stepping behavior for people post-stroke. Individuals with stroke may respond differently to BWS, so directly quantifying the behavior in this population is merited. In addition, as BWS is a common rehabilitation tool for other clinical populations that have difficulty controlling frontal plane stability, such as people with incomplete spinal cord injury [124] or lower extremity amputations [125]. These populations may have unique stepping responses to BWS. Individuals with incomplete spinal cord injury use a non-specific stepping strategy that prioritizes wide steps and increased double support time to maintain gait stability [34], [126]. Similarly, individuals with amputations use a non-specific strategy that also exhibits wide steps [127]–[129](Refs), increased step width variability [129], [130], and uses compensatory movements [127], such as hip circumduction and hip hiking, during the swing-phase of gait to maintain stability. As such, investigating population-specific responses to BWS is warranted.

The methods presented in this work can be used to characterize how any gait training interventions affect the control of stability. This dissertation demonstrates how a particular BWS system changes frontal plane stepping behavior but there are many types of BWS which likely produce unique walking dynamics. For perspective, BWS can have different methods of actuation (active vs passive), type of overhead fixation (0-3 translation degrees of freedom), and walking mode (treadmill vs overground). Active BWS systems dynamically adjust the vertical support force to better track the vertical motion of the COM, whereas passive systems provide a constant vertical force that are more restrictive. The simplest type of BWS is attached to a single fixed-point overhead, which limits translation in the anteroposterior and mediolateral directions if the system is active or all translation degrees of freedom if passive. More sophisticated systems, such as the BWS system that I used in this dissertation [47], use an overhead trolley system to follow the participant as the move forward and backward on the treadmill or overground. New BWS systems can provide full three dimensional tracking of the participant, which can dynamically scale the support force in any direction, and has been shown to better reproduce the dynamics of normal walking [67]. In addition, gait stability requirements are different for overground and treadmill walking [131], with the former being a more challenging dynamic environment. These various BWS devices will each apply a unique pattern of supporting forces to the body during walking, and each system can be directly evaluated to determine how they affect gait stability. Future work should explore how these various BWS devices interact with and affect the control of gait stability.

In Chapter 1, I proposed developing stability-based interventions that specifically target the control of gait stability. The intervention also needs to appropriately challenge the control of stability, which can be accomplished by scaling the lateral supporting force to the patient's abilities. For individuals who have severe gait deficits and cannot maintain their balance independently, an intervention can act to 'stabilize' the patient and decrease the need to actively maintain stability. The Damping Field is an example of this type of field, and the gain of the field can be scaled to the stability requirements of the patient and can be used to gradually increase the challenge of walking. For the moderately impaired who use compensations or non-specific strategies, a 'stabilizing' field can allow for practice of new gait patterns or more appropriate stepping strategies. Interventions can also be designed to make it more difficult to control gait stability, by applying perturbations or 'destabilizing' force fields. For instance, rather than damping the lateral movement of the COM, a force field can amplify or exaggerate movement of the COM. My laboratory has demonstrated that this field encourages practice of desirable stepping strategies for individuals with and without spinal cord injury that increase coordination between the COM and foot placement [109]. A potential approach for post-stroke gait training could be to combine these fields in a gait training regimen and determine if they lead to improvements in balance.

In my work, I measure how people adapt their steady-state stepping behavior when exposed to a new dynamic environment, but it is unclear if these new stepping strategies are 'learned' by the locomotor system. There is strong evidence that the central nervous system can form new representations between limb and body state, and this ability appears to be retained after a stroke. For instance, people post-stroke can learn more symmetric gait patterns with split-belt treadmill training [132]–[134], and these symmetry improvements can carry over to overground ambulation [132]. Recent work has demonstrated that individuals with stroke exhibit increased coordination between the COM and foot placement when exposed to lateral perturbations of the swing foot [42]. It remains unclear if lateral assistive forces can produce similar persistent changes in lateral stepping behavior for people post-stroke. Future studies can explore if people post-stroke can learn new stepping strategies when exposed to force fields that perturb the COM.

While it is likely that individuals with stroke can learn how to better coordinate their COM state and the lateral foot placements, it is unclear if this change would directly lead to improved walking balance or a reduction in falls. My work assumes that increased coordination between the COM and foot placement would result in a more 'appropriate' or 'better' stabilization strategy for real-world walking and that people post-stroke would learn a stepping strategy that is commonly used by healthy adults. However, this should not be assumed without evidence. There is a persistent debate in how to best approach post-stroke gait rehabilitation, and if a return to ‘normal’ behavior is the most desirable or preferable outcome. While it is possible that improved stepping coordination will lead to improvements in walking balance during community ambulation, future studies should evaluate how gait training with lateral assistive forces affects the incidence of falls and standard clinical outcome measures for balance.

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