Postural Control in Standing and Walking: Integrating Lower Extremity Force and Kinematic Factors

By

Jenny Bartloff

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The dissertation is approved by the following members for the Final Oral Committee:

Kreg Gruben, Professor, Kinesiology and Mechanical Engineering

Peter Adamczyk, Associate Professor, Mechanical Engineering

Christa Wille, Teaching Faculty, Research Analyst, Biomedical Engineering
and Center for Innovation in Engineering Education

Darryl Thelen, Professor, Mechanical Engineering and Biomedical Engineering

Josh Roth, Assistant Professor, Mechanical Engineering

David Brown, Senior Vice President and Dean,
University of Texas Medal Branch School of Health Professions

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Abstract

This dissertation investigates the neuromechanical principles underlying postural control and gait in individuals with neurologic impairments, aiming to enhance clinical assessment and rehabilitation. Through detailed analysis of novel metrics such as the Intersection Point (IP), Xi, and application of a novel method to analyze position and timing of minimum Foot Clearance (mFC), this research provides new insights into motor control strategies during standing and walking. The IP metric revealed distinct motor control patterns in post-stroke hemiplegia, while Xi demonstrated independent limb regulation during gait perturbations in healthy individuals. The novel mFC method, using IMUs and 3D scans, effectively evaluated foot clearance in individuals with Multiple Sclerosis who present with foot drop, comparing the efficacy of carbon-fiber ankle-foot orthoses and functional electrical stimulators as methods for avoiding foot contact during the swing phase of gait. It was then applied to understand where mFC occurs both spatially on the foot and temporally during swing phase to better understand how foot drop interventions alter risk of experiencing unwanted ground contact.

The findings underscore the potential of these approaches for clinical application, addressing gaps in our ability to comprehensively view motor control impairments. Future work will focus on achieving an understanding of the causes and consequences of changes in IP and Xi location, further establishment of their psychometric properties, development of these metrics as training targets in postural control interventions, a deeper exploration of the new horizon of mFC assessments enabled by the novel IMU/3D scan method, and analysis of real-world, remote monitored data from this

technique. The research included in this dissertation represents a step towards improving functional mobility and quality of life for individuals with neurologic injuries through targeted, evidence-based interventions.

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Chapter 1: Introduction

As a physical therapist with a specialty in neurologic practice, the choice to return to graduate school for PhD training in mechanical engineering came with a strong hope to impact my field more than is possible through clinical practice alone. It was clear to me that training as a scientist and engineer would give me the opportunity to contribute to broader change across the discipline of physical rehabilitation. During my time at University of Wisconsin-Madison (UW), I had the fortune to explore diverse aspects of engineering and engage in a multitude of layers of clinical and basic science research that all have potential to spark change to health care practice.

In my work with the Neuromuscular Coordination Lab (NCL), I gained a unique perspective on postural control in standing and stance-phase of walking through consideration of the underlying, requisite mechanical problems which the intact adult central nervous system (CNS) has solved. By studying how humans control the direction of the forces between the foot and ground, our lab's research provides additional perspectives to previous studies of motor control impairment related to postural control in neurologically impaired populations, such as those with post-stroke hemiplegia. In physical medicine and rehabilitation, it remains common practice to treat easily observable manifestations of central nervous system (CNS) damage-induced motor control impairment, such as muscle weakness and spasticity (Beyaert et al., 2015; Raghavan, 2022; Wist et al., 2016). However, evidence is mixed on whether improvements in these domains lead to significant enhancements in quality of life, gait speed, or other functional measures (Selves et al., 2020; Suputtitada et al., 2024).

Further, treatments focused on mechanical features of balance impairment, such as aspects of the center of pressure (CP), including but not limited to CP excursion and velocity, have also failed to yield significant improvements in functional and quality of life domains (Geurts et al., 2005).

From a mechanical engineering perspective, adults with post-stroke hemiplegia transition from a neuromuscular control system that efficiently manages upright postural balance pre-stroke to one that solves these mechanics with altered, less effective strategies. While the new strategies might in some cases be “good enough” for preserving upright mobility, post-stroke neuromuscular coordination is typically less efficient, less effective, and slower. Effective treatment plans for restoration of pre-stroke coordination must be based on an understanding of the mechanical requirements for postural stability, including strategies used in healthy individuals and those with neurologic impairment. For example, center of pressure measures provide a partial understanding of standing mechanics, but they fall short of fully describing the mechanics of the task of standing. In the NCL, we address this gap specifically by pairing CP movement with the direction of the foot-on-ground force (θF), a necessary element to control both linear and angular momentum. This component (θF) is often completely overlooked in studies of motor control of postural stability in standing and walking.

Seeing the potential of this combinational approach to augment the understanding of motor control within the field of neurorehabilitation, and to develop more effective treatments for impairments of motor control in those with neurologic injury, I developed two goals for my time in the NCL: 1) propose a more comprehensive theoretical understanding of how θF-related metrics as they relate to motor control in both health and neurologic impairment and 2) Facilitate future clinical implementation of these measures. Toward those goals I quantified how individuals in health and with post-stroke hemiplegia adapt their postural control strategy for both lower extremities to speed and posterior force perturbations during walking. This analysis used a θF-related stance-phase walking metric, called Xi, which combines θF with its point of application to describe control of upright posture. I also designed and implemented empirical studies of standing postural control in individuals with post-stroke hemiplegia and analyzed psychometric properties of our θF-related standing metric, the intersection point (IP).

In collaboration with Dr. David Brown at University of Texas Medical Branch, we furthered our understanding of the Xi strategy of postural control during the walking task in response to symmetric (posterior force applied at the hips) and asymmetric (a split-belt speed change under one leg) disturbances. This study represented the first application of Xi to characterize motor control of walking’s stance phase in the face of perturbations. It has helped to establish Xi as a metric that is able to detect changes in stance-phase postural control and has brought to light important questions about this Xi and its related underlying mechanics worthy of further study.

Through these projects at the NCL, we have successfully improved understanding of how the Xi strategy is altered in healthy adults under perturbations of speed and posterior external force during walking. We have assisted in establishing good validity and reliability of the IP for measurement of standing postural control. The establishment of psychometric properties is a crucial step in evaluating any new measure for its continued utility in research and potential for translation to the clinic. We plan to build on this work with further exploration into Xi strategy in a cohort of individuals with post-stroke hemiplegia, and a further exploration of the test-retest reliability of the IP; both studies now stand at the data analysis stage. Further, through our published study of IP strategy in individuals with post-stroke hemiplegia and healthy controls, we established: 1) that while the post-stroke population still demonstrates presence of an IP for both legs, 2) the IP location of paretic and non-paretic limbs differs from eachother and from healthy controls, and 3) the IP is a viable outcome measure for assessing disruption to motor control strategy.

During my time at UW, I aimed to gain a wide breadth of skills in mechanical engineering principles to enhance my understanding of biomechanics, specifically in the domains of data collection and analysis. Beyond studying standing and gait stance phase, I wanted to acquire tools for exploring the complete gait cycle. With the Real Life Wearables (RLW) team at the Biomechatronics, Assistive Devices, Gait Engineering and Rehabilitation Laboratory (BADGER Lab), we worked to develop and apply a novel method for determining whole-foot ground clearance. By applying this method to a population of individuals with impaired ankle control secondary to CNS damage (specifically, foot drop consequent to Multiple Sclerosis, MS), we aimed to provide new insights into how two commonly used, but mechanically distinct, treatment approaches impact foot positioning with respect to the ground and the risk of unwanted ground contact and falls.

We successfully developed this novel method using a single inertial measurement unit (IMU) on the foot, paired with a 3D scan of the user's foot and shoe. This work is currently in press. We have also applied this method in users without CNS damage but with a below knee amputation. In the CNS-impaired population, we have successfully begun to explore how this method can be leveraged to provide a greater understanding of the effectiveness of varied treatments for ankle control impairment. By analyzing the location where the point of minimum foot clearance occurs during swing phase both spatially (where on the shoe) and temporally (where in time during swing phase), we are developing a unique view of foot kinematics that was previously unexplored and with our novel method, has the potential for inexpensive, broad clinical use.

The research presented in this dissertation has provided significant insights into the mechanical underpinnings of postural control and the impact of neurologic impairments on gait dynamics. Through improving understanding of the Xi strategy, and through validation of the intersection point (IP), we have advanced the understanding of how humans control foot-ground forces during both standing and walking. These metrics have shown promise in explaining motor control impairments beyond traditional measures, offering new perspectives on the mechanical requirements for postural stability.

By combining engineering principles with clinical insights with the RLW team at the BADGER Lab, we have developed a new method of monitoring gait and evaluating 3D foot movement which facilitates unique insight into risk of falls in populations with gait impairment. Further progress on this work has potential to contribute to the development of more effective treatment strategies in gait rehabilitation.

Building on these advances, this dissertation highlights many areas that require further exploration. The relationship between mechanical measures of postural control and functional outcomes needs to be better understood to fully translate the θF-related metrics into clinical practice. Specifically, further exploration is needed into the meaning of changes in these metrics and their sensitivity to changes in neuromuscular dysfunction. Further down the line of clinical translation, development of θF-related interventions will require extensive care to ensure their greatest possibility of success in clinical trials. Finally, the long-term effects of such interventions on gait stability and their integration into routine rehabilitation protocols will require establishment.

Further, the RLW team's novel method for estimation of whole foot clearance has been established as useful and valid, but my dissertation work has only begun to explore the power of this method to enhance assessment of gait interventions for swing-phase dysfunction. Applying this method in broader populations and on a larger scale will be needed to fully establish this analytical approach as clinically useful.

Overall, this research has made meaningful contributions to the field of neurorehabilitation by enhancing our understanding of the biomechanics of postural control and gait. It utilizes the potential of interdisciplinary approaches (mechanical engineering, neurologic rehabilitation, biomechanics) to address complex clinical challenges, and paves the way for future studies aimed at improving the quality of movement and of life for individuals with neurologic injuries through targeted, evidence-based interventions.

Chapter 2: Background

Despite advances in neurorehabilitation, individuals dealing with postural control impairments face persistent imbalance and an elevated risk of falling (Pin et al., 2024). The problem of postural imbalance impacts activities of daily living, often leading to reduced mobility independence and quality of life. **The problem involves both the inherent balance impairments post-neurological injury and the challenges that current rehabilitation practices face in comprehensively assessing and treating this problem (Piscitelli, 2016; Reinkensmeyer & Dietz, 2016)**. While strides have been made in understanding and rehabilitating impairments in postural control, the high prevalence of imbalance, fear of falling, and reduced independence with upright mobility for individuals in neurologically impaired populations indicates a gap between our knowledge and its application in creating safe, stable mobility.

Addressing the issue of postural control impairment is not just about preventing falls and physical injury; it's about preserving independent mobility, confidence, and quality of life. The perception of imbalance can lead to a vicious cycle of fear, activity avoidance, and subsequent physical decline. Fear of falling is itself a significant risk factor for a fall (Pin et al., 2024). This emotional and psychological toll, along with the economic burden of medical interventions due to falls, **underscores the critical need for a focused and effective approach to improve postural control in individuals with neurologic impairments.**

Unfortunately, the effectiveness of conventional neurorehabilitation therapies for restoring gait and balance post neurologic injury has been modest, partly attributed to the absence of standardized assessments and a robust foundation of empirical evidence (Reinkensmeyer & Dietz, 2016; Stinear et al., 2020). Improvement in clinical trials focusing on motor function such as postural control can be achieved by carefully selecting specific, hypothesis-driven endpoint measures (Stinear et al., 2020). To address the problems of postural control impairment facing neurologic populations, we need endpoints that allow full characterization of the impairment.

Postural control impairment, or instability, is multi-faceted and difficult to capture in a single outcome measure. To successfully maintain upright posture, the brain is simultaneously dealing with multiple categories of sensory input, commanding motor output, and working to minimize energy expenditure (Balbinot et al., 2020). There are many ways to categorize the tasks which underlie successful postural control. Fay Horak et. al. created the Balance Evaluation Systems Test (BESTest) (Horak et al., 2009) to comprehensively evaluate six systems which underlie postural balance control: Biomechanical Constraints, Stability Limits/Verticality, Anticipatory Postural Adjustments, Postural Responses, Sensory Orientation, and Stability in Gait. This approach aims to identify which system(s) may benefit from intervention to treat problems underlying balance deficits.

Ultimately, successful postural control is a reflection of how the CNS solves the mechanics of upright standing. When viewed as a mechanical task, the final common output of the nervous system's inputs and control strategy is the orientation, magnitude, and location of the force between the ground and feet, or foot force (F). F's location (center of pressure, or CP) and orientation (θ F) at any given moment are fine-tuned by alterations in relative lower extremity joint torques, and dictate the torque created about the whole body center of mass. This is true during quiet standing and the stance phase of gait. During swing phase, the relative lower extremity joint torques over time determine the joint angles which determine foot location and thus ground clearance and foot location at touch-down.

The unifying approach to the included studies is the experimental manipulation of postural stability and examination of resultant responses in terms of metrics which are designed to capture the above biomechanical features of postural control. Successfully maintaining postural control through each of these stages of upright mobility requires effective CNS management of lower extremity joint torques and thereby dictates how the foot will or will not interact with the walking surface from a kinetic standpoint, to manage torque about the center of mass and remain upright.

Intersection Point: Assessment of Standing Postural Control

Building on the recognition of the limitations inherent in current neurorehabilitation approaches, this dissertation research characterizes the postural control of human standing and stance phase of walking by examining the modulation of foot-on-ground force (F). The relationship between orientation and center-of-pressure (CP) of F serves as a critical interface between motor intention and output, offering a window into the precise coordination of lower extremity muscle activity to achieve upright mobility tasks. When observing the sagittal plane during a period of quiet standing, the CP moves fore and aft, and as it does, F's lines-of-action change orientation. These lines-of-action pass near an intersection point (IP) within discrete frequency bands. Previous experiments have shown the *height* of the intersection point (zIP) to be responsive to various balance challenges (Dutt-Mazumder & Gruben, 2021a; Yamagata et al., 2021a). zIP provides a single metric capable of describing the motor control system's control of torque about the CM. Gaining an understanding of zIP alterations in states of neurologic injury therefore provides crucial information for developing novel treatment strategies rooted in the mechanics and postural control required for the task.

Xi: Assessment of Stance-Phase Postural Control During Gait

Recognizing the importance of zIP in exploring typical and disordered neuromuscular strategies during quiet standing, this dissertation extends to analyze locomotion, the dynamic counterpart of postural stability. Observing the intersection of F vectors during the stance phase of gait, previous work from the Neuromuscular Coordination Laboratory (NCL) defined a metric of postural control labeled Xi. Like zIP, the Xi metric provides a perspective into the coordination of lower extremity muscles to maintain balance and resist angular perturbations. Despite the ability of Xi to describe a component of F variation during non-disabled walking, the healthy brain's ability to adapt Xi strategy has not been explored. A goal of this dissertation is to complete an inquiry into how Xi may differ between dominant and non-dominant limbs and under speed and balance perturbations. Ultimately, both Xi and zIP leverage a unique understanding of the interaction of motor control with body mechanics for creation of new clinical assessment and training tools for addressing neurologically impaired gait and balance.

Minimum Foot Clearance:
Novel Methods for Assessing Swing Phase Foot Kinematics

In line with the aims of this dissertation to assess postural control in standing and walking through the lens of task mechanics, the final aspect focuses on adopting a pioneering gait analysis method developed in UW-Madison's BADGER Lab which utilizes a single inertial measurement unit (IMU) paired with a 3D scan of the user's foot and shoe. This method enhances assessment of foot clearance during swing phase, a period susceptible to tripping and falls. Current foot clearance analysis methods, often limited to monitoring a single point on the shoe, fail to capture the complex 3-dimensional foot kinematics present in overground walking.

Minimum Foot Clearance:
Development of Novel Analyses for Understanding Effectiveness of Gait Interventions

By integrating a single IMU with a personalized 3D shoe scan, this approach offers a comprehensive analysis of minimum foot clearance (mFC), detailing the position of the foot relative to the ground through swing. This facilitates an understanding of how relative lower extremity joint torques may differ in states of altered postural control to change the location of the point of minimum clearance spatially (where on the shoe), and temporally (when in time during swing phase). My role in this project was to assist in developing this novel methodology and to interpret the outcomes from this advanced method when applied to a cohort of persons with MS (PwMS), aiming to provide valuable insights that could reshape gait analysis and intervention strategies for the condition of foot drop, where ankle muscle control is impaired, often leading to scuffing (unintended foot-ground contact) and an elevated fall risk.

Innovation

The overall goal of this dissertation was to adopt three distinct methodologies, each offering a robust description of whole-body control in standing and walking tasks with a single metric. The main goals of physical rehabilitation clinicians treating patients with neurologically induced gait impairments are to maximize functional ability and independence while minimizing risk of harm such as falls. Crucial to this process is the reliance on objective biomechanically-sound assessments to guide treatment and monitor progress. Such measures are also crucial to monitoring progress and determining treatment efficacy. However, existing measures often provide a fragmented or superficial understanding of gait and balance mechanics, limiting their utility in clinical settings.

The work presented in this dissertation demonstrates that zIP and mFC location are able to represent the complexities of the mechanical task of postural control in one single metric for their respective component of upright mobility. These metrics show promise as future clinical analysis tools, but stand at relatively early stages in their path from engineering design toward clinical translation. While Xi may be less feasible to implement clinically due to high equipment costs, my dissertation work demonstrates that it provides valuable insight into motor control during the stance phase of gait, warranting further research. By providing a more complete picture of postural control in each of these component tasks of upright mobility, zIP and mFC could lead to more effective clinical assessments, treatments, and ultimately, better patient outcomes, while Xi, though less feasible for clinical use, offers valuable insights that merit further research.

Chapter 3: Frequency-dependent behavior of paretic and non-paretic leg force during standing post stroke

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Abstract

Maintaining upright posture in quiet standing is an important skill that is often disrupted by stroke. Despite extensive study of human standing, current understanding is incomplete regarding the muscle coordination strategies that produce the ground-on-foot force (F) that regulates translational and rotational accelerations of the body. Even less is understood about how stroke disrupts that coordination. Humans produce sagittal plane variations in the location (center of pressure, xCP) and orientation (Fx/Fz) of F that, along with the force of gravity, produce sagittal plane body motions. As F changes during quiet standing there is a strong correlation between the xCP and Fx/Fz time-varying signals within narrow frequency bands. The slope of the correlation varies systematically with frequency in non-disabled populations, is sensitive to changes in both environmental and neuromuscular control factors, and emerges from the interaction of body mechanics and neural control. This study characterized the xCP versus Fx/Fz relationship as frequency-dependent Intersection Point (IP) heights for the paretic and non-paretic legs of individuals with history of a stroke (n=12) as well as in both legs of non-disabled controls (n=22) to reveal distinguishing motor coordination patterns. No inter-leg difference of IP height was present in the control group. The paretic leg IP height was lower than the non-paretic, and differences from control legs were in opposite directions. These results quantify disrupted coordination that may

characterize the paretic leg balance deficit and non-paretic leg compensatory behavior, providing a means of monitoring balance impairment and a target for therapeutic interventions.

Introduction

Stroke is the leading cause of serious long-term disability in the United States, with a current population of approximately 795,000 survivors (Tsao et al., 2022). Roughly 15-30% live with permanent disability (Yao et al., 2021). Despite rehabilitation efforts, many stroke survivors continue to experience imbalance and significantly elevated fall risk (Weerdesteyn et al., 2008) associated with impaired nervous system responses to environmental and internal perturbations (Garland et al., 1997; Pollock et al., 2017). Effective balance therapy requires understanding of typical motor control strategies and post-stroke disruptions. Unfortunately, the relationship between neuromuscular mechanisms and standing balance behavior is poorly understood. Despite the essential nature of balance to daily living tasks (Schmid et al., 2013), the evidence supporting physical therapy approaches is limited (Arienti et al., 2019; Geurts et al., 2005; Pollock et al., 2007).

Various observational and training studies involving quiet and perturbed standing show distinguishing aspects of net ground-on-foot force (F) and its location of application (center of pressure, CP) between individuals with stroke-related motor impairment and non-impaired individuals (Corriveau et al., 2004; Geurts et al., 2005; Karlsson and Frykberg, 2000; Niam et al., 1999; Palmieri et al., 2002; Pyöriä et al., 2004). During quiet standing, post-stroke hemiparetic patients commonly present with weight-bearing asymmetry favoring larger non-paretic vertical force magnitude (de Haart et al., 2004; Roerdink et al., 2009; Sackley, 1991; Schinkel-Ivy et al., 2016). Increased magnitude and variability in whole-body center of mass (CM) excursion and combined-feet CP excursion and velocity are also commonly observed (de Haart et al., 2004; Sackley, 1991). The differences in these variables between non-impaired individuals and those post-stroke are usually considered deficits and used as recovery metrics. In addition, the severity of deviation from neurologically intact individuals is often inversely correlated with measures of functional balance (Blum and Korner-Bitensky, 2008; Corriveau et al., 2004; Karlsson and Frykberg, 2000; Pyöriä et al., 2004). Training studies have focused on restoring post-stroke CP and CM metrics toward that of neurologically intact individuals. However, that approach yields mixed results (Geurts et al., 2005). Further, the relationship between changes in these measures and practical improvements in activities of daily living is not conclusive (de Haart et al., 2004; Langhorne et al., 2009; Niam et al., 1999). These gaps further indicate the need for a deeper understanding of the mechanisms of standing balance and its impairment.

A shortcoming of the above-mentioned assessments is that they do not fully account for variations in the torque of F about the CM, which controls rotational motion of the body. The torque of F about the CM, which drives whole-body rotational acceleration, is a function of the relative locations of the CP and CM as well as the magnitude of F and orientation of F (represented as the ratio of anterior-posterior to vertical force, Fx/Fz, Fig. 3.1). A balance metric sufficient to characterize sagittal plane standing control must reflect the control of the whole-body translational *and* rotational accelerations and thus must include both anterior-posterior CP (xCP) *and* Fx/Fz.

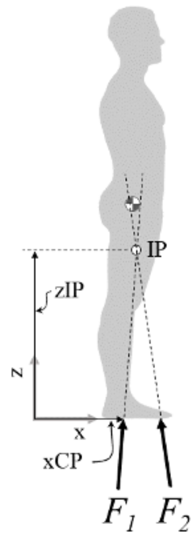

Figure 3.1: During quiet standing the force of the ground on a foot or feet (F) varies in location (xCP) and orientation (ratio Fx/Fz) across time (F_1, F_2). An emergent property of human motor control is that variations in F that occur within a narrow frequency band exhibit lines of action that tend to pass near an intersection point (IP) that is located a distance (zIP) above the support surface.

The few studies that addressed horizontal force during standing discuss its impact on translational control of the CM without consideration for its rotational contribution (Winter, 1995; Zatsiorsky and Duarte, 2000). The decomposition of CP variation into two frequency components showed distinct behaviors for a low frequency component (< ~ 0.4 Hz, termed rambling) and a high frequency component (> 0.4 Hz, termed trembling) (Zatsiorsky and Duarte, 2000, 1999). The rambling component was described as a moving reference point in the plane of the support surface and the trembling component was considered an oscillation of control around that reference point. In the trembling component the horizontal component of F was found to be directed toward, and moderately correlated with, the deviation of the CP from the rambling reference point. The authors noted the beneficial nature of this pattern in that such a horizontal F translationally accelerates the CM toward the reference point, aiding in the restoration of deviations from upright posture.

Upon closer inspection of the Fx and xCP variations in the trembling frequency range (>0.4 Hz) it becomes apparent that Fx/Fz exhibits strong covariation with xCP within narrow frequency bands (Fig. 3.2) (Boehm et al., 2019). That observation considered the CP location expressed in a laboratory-fixed reference frame. However, to understand the rotational acceleration effect of F on the whole body, the F and thus the CP must be expressed relative to the CM which moves anterior-posterior relative to the floor with body sway. Expressing xCP as the CP location relative to a vertical line through the CM, a perfect linear covariation of Fx/Fz with xCP would result in all F lines-of-action passing through a single intersection point in a reference frame attached to the CM (IP, Fig. 3.1). The height of that IP (zIP) within each frequency band is estimated by the slope of the first principal component of the Fx/Fz vs xCP relationship (Fig. 3.2). Steeper slope of the Fx/Fz vs xCP covariation is equivalent to a zIP closer to the floor. The measured F lines-of-action do not all exactly intersect at a point but tend to pass close to the IP. Thus, a line from the IP to the instantaneous xCP closely predicts Fx/Fz. The zIP is a function of both Fx/Fz and xCP and therefore IP height is related to the torque imposed by F about the whole-body CM. That torque drives whole-body rotational acceleration and thus zIP is related to the rotational contribution of F to standing control. The Fx portion of Fx/Fz drives translational (horizontal) acceleration of the whole body CM. Variation in xCP, Fx, and Fz are primarily produced by varying

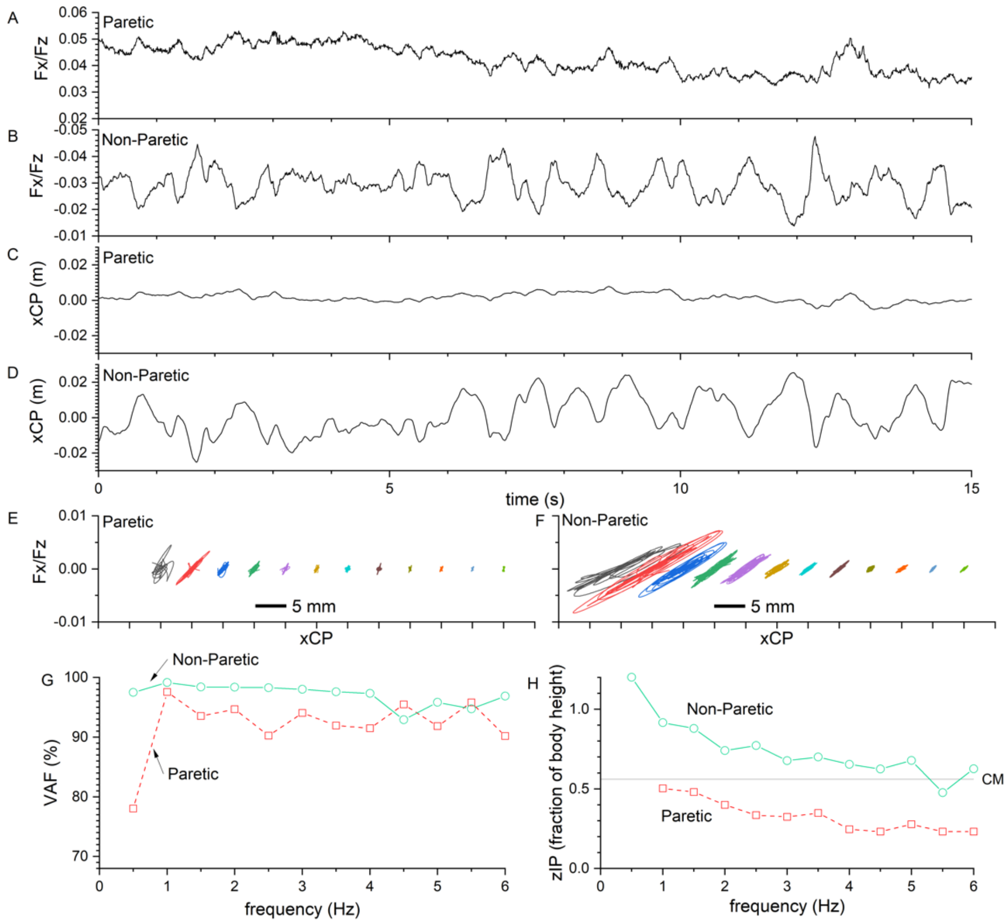

Figure 3.2: Sample data from a typical individual (stroke group) illustrate steps in the process of converting the Fx/Fz (panels A & B) and xCP (panels C & D) time series into the frequency-dependent zIP (panel H). The relationship between the band-pass filtered Fx/Fz and xCP time series signals (panels E & F, each frequency band is offset to the right by 0.005 for visualization) exhibits a high correlation consistent with the F lines of action passing near an intersection point (Fig. 3.1). For a given frequency band, the high correlation between the Fx/Fz and xCP filtered signals is reflected in the VAF >90% (panel G). For each frequency band the inverse slope of the first principal component of the Fx/Fz versus xCP data (panels E & F) yields the IP height (zIP, panel H) which decreased with increasing frequency.

lower extremity joint torques (hip, knee, and ankle), though the joints do not contribute equally to each F component (Hof, 2001). Thus, zIP provides insight into the multi-joint coordination chosen by humans to drive the translational *and* rotational body motions of standing (Boehm et al., 2019).

The potential for zIP analysis to provide insight on motor control is supported by optimal control modeling that showed dynamic mechanical coupling between the primary body segments does not fully account for the emergent IP behavior, but that motor control strategies contribute to the typical zIP vs frequency curve (Shiozawa et al., 2021). Additional investigations that probed IP sensitivity to the motor control contribution demonstrated that whole-body muscle co-activation shifted the location of the curve (Yamagata et al., 2021), and sloped standing surfaces changed its shape (Dutt-Mazumder and Gruben, 2021). It remains unknown whether or how the IP behavior changes following the neural disruption caused by stroke, but this previous work suggests potential for the IP approach to both identify signature post-stroke behavior and guide its recovery. Distinct implications for balance could result from a shift or shape change in the curve, and absence of IP behavior would suggest an entirely different form of disruption.

Therefore, this study assessed the presence of an IP behavior following stroke, used zIP analysis to characterize the frequency-dependent F generated by paretic and non-paretic legs in humans post-stroke, and then compared those post-stroke results with those of non-disabled control participants. While previous published reports assessed zIP for both legs combined, we hypothesized that control participants would exhibit similar zIP in each leg based on pilot research showing IP behavior in individual legs. Also, we hypothesized that an IP behavior would be present in both paretic and non-paretic legs, and that the paretic and non-paretic zIP would differ from each other and from controls due to stroke-induced alterations from typical control (Rogers et al., 2004) (details in Discussion). The presence of IP behavior post-stroke would reveal preservation of some portion of typical muscle coordination, while inter-leg zIP differences and deviations from control can be used to characterize the specific disruptions to translational and rotational stabilization of upright standing posture and aid in identifying abnormal muscle coordination patterns that should be the target of rehabilitation.

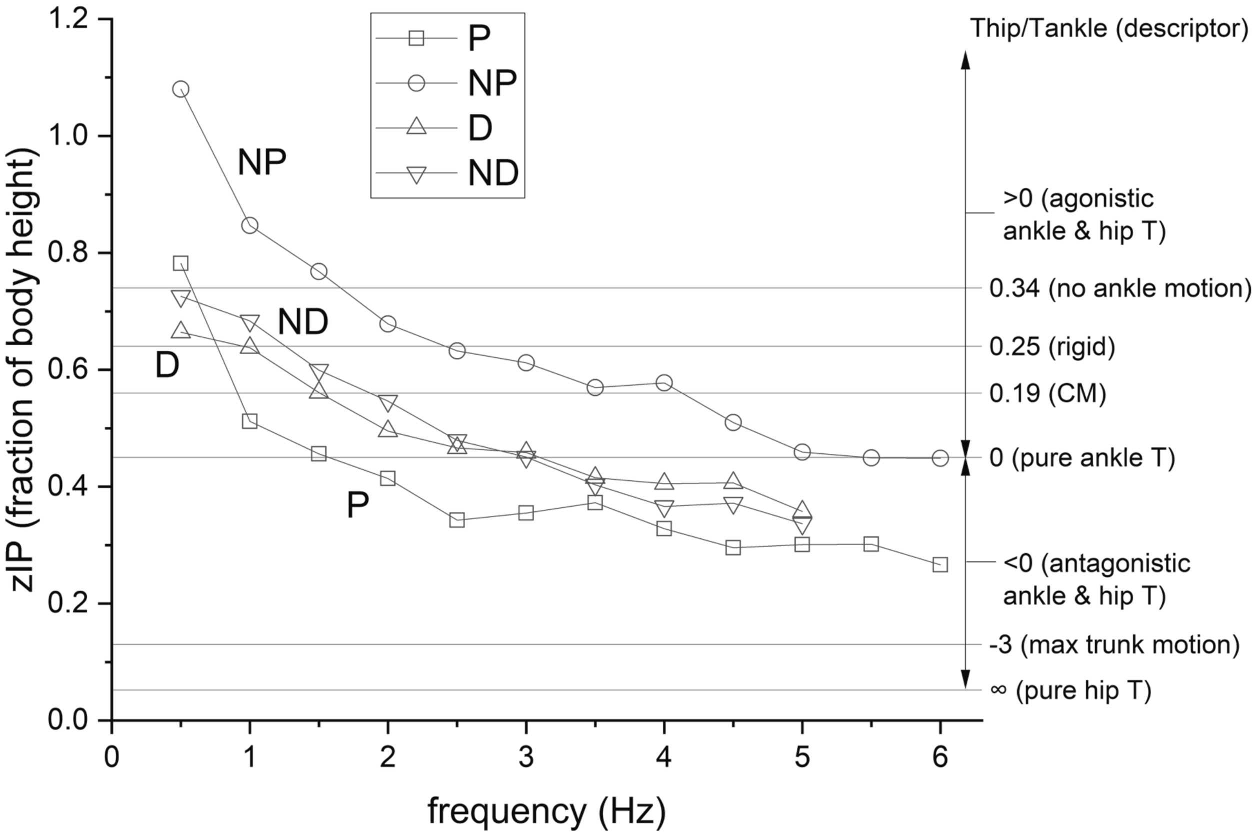

Fig. 3.3: For all legs, the mean zIP varied with frequency in a similar manner with the zIP decreasing with increasing frequency. The zIP in the control group did not differ between the dominant (D) and non-dominant (ND) legs (Table 3.1, Fig. 3.4). However, in the stroke group the paretic (P) leg zIP was lower than in the non-paretic (NP) leg. Compared to the control group at each frequency, the zIP of the paretic leg was typically lower and the non-paretic leg was higher. An anthropometric two segment inverted-pendulum model driven with proportional hip and ankle torques (Appendix A) produced IP's at various heights depending on the ratio of hip/ankle torque. The zIP for pure ankle torque is below the CM and the zIP for rigid body motion is above the CM. Other special cases are also noted.

Methods

Participants & Data Collection

Participants (n=12, age=62±14 years mean ± standard deviation, 4 female) with chronic hemiplegia (>6 months post-stroke) volunteered. Inclusion criteria: ability to stand for five minutes without assistance and without pain, walk 10 meters with or without assistance. Exclusion criteria: recent history of chronic falling (>1 in the past month), history of additional neurologic disease, >1 stroke. Participants were not selected by type or location of stroke. For comparison, a control group of unimpaired adults was included (n=22, age=69±11 years, 11 female) (dos Santos et al., 2017). The protocols were approved by local Institutional Review Boards and written informed consent was obtained from all participants prior to enrollment.

Participants stood quietly, arms at sides, gazing at an eye-height target, with a force plate under each foot for the stroke and control groups (stroke force plates (Boehm and Gruben, 2018), control force plates (dos Santos et al., 2017)). The CP and 3-dimensional force of the ground on the feet (F) were collected at 100 Hz. Only the sagittal plane F was analyzed (anterior-posterior Fx & xCP, vertical Fz). Data collection duration was: 15s for nine stroke participants, 60s for three stroke participants, and three 60s trials for each participant in the control group. Previous analyses (unpublished) have shown that the zIP curves have similar shape for standing durations of ≥10 s justifying our use of the full data collection duration from each study despite the duration difference.

Analysis
The frequency content above ~0.4 Hz was considered due to the previously reported correlation between xCP and horizontal force for the trembling frequency component (Boehm et al., 2019; Zatsiorsky and Duarte, 2000, 1999). The variation in anterior-posterior CM location was calculated by double integrating Fx/m with respect to time (m = body mass). To reduce error accumulation associated with integration due to errors in Fx measurement, the Fx/m was high-pass filtered at 0.006 Hz before double integration, based on the assumption of bounded CM motion during standing. This approach was validated by comparing the zIP calculated with xCP expressed relative to xCM from two sources, one with xCM measured from measured body kinematics (motion capture) and the other with xCM calculated from Fx as described above. The 0.06 Hz threshold was chosen to minimize difference between the two xCM sources. The difference in zIP between these two methods had a mean absolute difference of 1.1% across the frequencies of 1, 2, 3, and 4 Hz, with the largest mean absolute difference being 3.1% at 1 Hz. To resolve the differences in IP height by frequency, the Fx/Fz and xCP time series were band-pass filtered (2^{nd}-order zero-lag Butterworth) using 12 evenly-spaced bands with center frequencies from 0.5 to 6 Hz (every 0.5 Hz) with bandwidths of 0.5 Hz.

For each standing trial and each leg, the zIP was calculated for each frequency band. Principal component analysis characterized the primary covariation between Fx/Fz versus xCP (Fig. 3.2) (Boehm et al., 2019). The zIP was the inverse slope of the 1^{st} principal component and expressed as a fraction of body height. The horizontal location of the IP would be reflected by the distance of the Fx/Fz vs xCP line from the origin, however the bandpass filter removes the DC component so the IP location only has a vertical dimension (zIP). The variance-accounted-for (VAF) of the 1st principal component described how well an IP described the covariation between Fx/Fz and xCP (Boehm et al., 2019). Within each frequency band a zIP value was calculated if VAF > 80%. For each frequency band with at least five zIP values (VAF>80%) the mean zIP for each leg was calculated (stroke paretic, stroke non-paretic, control dominant, control non-dominant).

To test for zIP and VAF population mean differences between legs we chose four frequencies (1, 2, 3, and 4 Hz) sufficient to describe the typical shape of the zIP vs frequency curve. As standing trials were acquired in triplicate for each control participant but only once for each stroke participant, we averaged the triplicate data zIP and VAF values at each frequency within each control participant to align with the single-trial data of the stroke participants. To assess differences between the four legs (stroke paretic, stroke non-paretic, control dominant, control non-dominant) we utilized mixed effects ANOVA models (R version 4.3.1) individually at each frequency of interest (1, 2, 3, 4 Hz). Each ANOVA model had a 4-level factor predictor of leg by group (paretic=P, non-paretic=NP, dominant=D, non-dominant=ND), and participant as a random effect to account for two measures per participant, one per leg. Post-hoc comparisons were made in R (version 4.3.1 2023-06-16 ucrt) utilizing emmeans (version 1.8.8) with Holm adjustment (Holm, 1979). All p-values were assessed at the 5% significance level.

The calculation of zIP involves the relationship between Fx, Fz, and xCP such that independent systematic variation in these variables should be evaluated for effect on zIP. For example, the paretic leg is typically found to have lower mean Fz and lower xCP variation than the non-paretic leg (de Haart et al., 2004). While Fz typically varies little at frequencies in the bands we evaluated (≥0.5 Hz), the mean Fz for a trial can differ widely from the typical value of near 50% of body weight, especially following stroke where the paretic Fz is typically much less than 50% (Roerdink et al., 2009). To assess the possibility that reduced Fz could lower zIP we quantified the correlation between zIP and mean Fz across all trials within each leg at each representative frequency using the Pearson correlation coefficient (r). We considered r<0.3 to be weak, 0.3<r<0.5 to be moderate, and >0.5 to be strong correlation. Similarly, a reduction in xCP variability could lower zIP and so we quantified the correlation between zIP and the standard deviation of xCP for each trial at each representative frequency.

1 Hz	NP	D	ND
P	<.0001 ***	0.0278 *	0.0041 **
NP		0.0004 ***	0.0051 **
D			0.2484
2 Hz	NP	D	ND
P	<.0001 ***	0.0938	0.0148 *
NP		0.0015 **	0.0213 *
D			0.1078
3 Hz	NP	D	ND
P	<.0001 ***	0.0347 *	0.0384 *
NP		0.0007 ***	0.0005 ***
D			0.8103
4 Hz	NP	D	ND
P	<.0001 ***	0.4331	0.5708
NP		0.0018 **	0.0002 ***
D			0.5708

Table 3.1: zIP p-values. At all frequencies the dominant (D) and non-dominant (ND) legs had similar zIP while the paretic (P) and non-paretic (NP) legs differed. Ranges of p-values are indicated by * <0.05, ** <0.01, *** <0.001.

1 Hz	NP	D	ND
P	0.0256 *	0.1677	0.4585
NP		0.7021	0.4585
D			0.7021
2 Hz	NP	D	ND
P	0.0006 ***	0.0802	0.2985
NP		0.2985	0.0707
D			0.2985
3 Hz	NP	D	ND
P	0.0264 *	0.6451	0.5964
NP		0.0102 *	0.0009 ***
D			0.5964
4 Hz	NP	D	ND
P	0.0688	0.0102 *	0.0004 ***
NP		<.0001 ***	<.0001 ***
D			0.0799

Table 3.2: VAF p-values: The ability of an intersection point to capture the covariation of xCP and Fx/Fz was reflected in the values of VAF that were mostly similar across legs at 1 and 2 Hz. At 3 Hz the non-paretic leg VAF differed from the other legs and at 4 Hz the paretic and non-paretic had similar VAF but differed from the dominant and non-dominant legs, which were similar to each other. Ranges of p-values are indicated by * <0.05, ** <0.01, *** <0.001.

Results

Within most frequency bands, the F measured during quiet standing exhibited an approximately linear relationship between the Fx/Fz and xCP (Fig. 3.2 E,F) for all groups and legs as indicated by the VAF being above 80% (Fig. 3.5). The ANOVA models revealed an effect of leg on VAF at all four frequencies ($df = 3$, $F>3.46$, $p<0.026$). Post-hoc comparisons revealed the following differences (Table 3.2 and Fig. 3.5). At 1, 2, and 3 Hz the VAF was consistently high (>85%) and similar across legs with the exception that the non-paretic leg was higher than the paretic and the non-paretic was higher than the other legs at 3 Hz. At 4 Hz the dominant and non-dominant legs had lower VAF than the paretic and non-paretic legs. The overall high VAF provides confidence that the IP behavior captures much of the variation in F and thus zIP is an informative summary of that behavior.

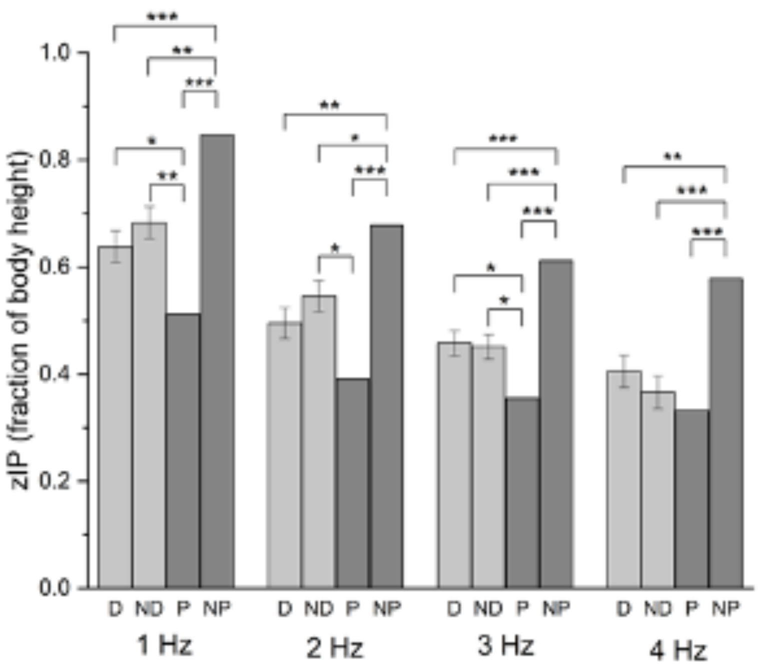

Figure 3.4: At all frequencies the mean zIP values of the dominant and non-dominant legs were similar to each other while the paretic (P) leg was lower than the non-paretic (NP). The paretic leg was typically lower and the non-paretic typically higher than the control participant legs. Error bars indicate ± standard error. The abscissa abbreviations are: D (Dominant control), ND (Non-Dominant control, P (Paretic stroke), NP (Non-Paretic stroke). Ranges of p-values are indicated by * <0.05, ** <0.01, *** <0.001.

1 Hz	NP	D	ND
P	0.0256 *	0.1677	0.4585
NP		0.7021	0.4585
D			0.7021

2 Hz	NP	D	ND
P	0.0006 ***	0.0802	0.2985
NP		0.2985	0.0707
D			0.2985

3 Hz	NP	D	ND
P	0.0264 *	0.6451	0.5964
NP		0.0102 *	0.0009 ***
D			0.5964

4 Hz	NP	D	ND
P	0.0688	0.0102 *	0.0004 ***
NP		<.0001 ***	<.0001 ***
D			0.0799

Table 3.3: VAF p-values: The ability of an intersection point to capture the covariation of xCP and Fx/Fz was reflected in the values of VAF that were mostly similar across legs at 1 and 2 Hz. At 3 Hz the non-paretic leg VAF differed from the other legs and at 4 Hz the paretic and non-paretic had similar VAF but differed from the dominant and non-dominant legs, which were similar to each other. Ranges of p-values are indicated by * <0.05, ** <0.01, *** <0.001.

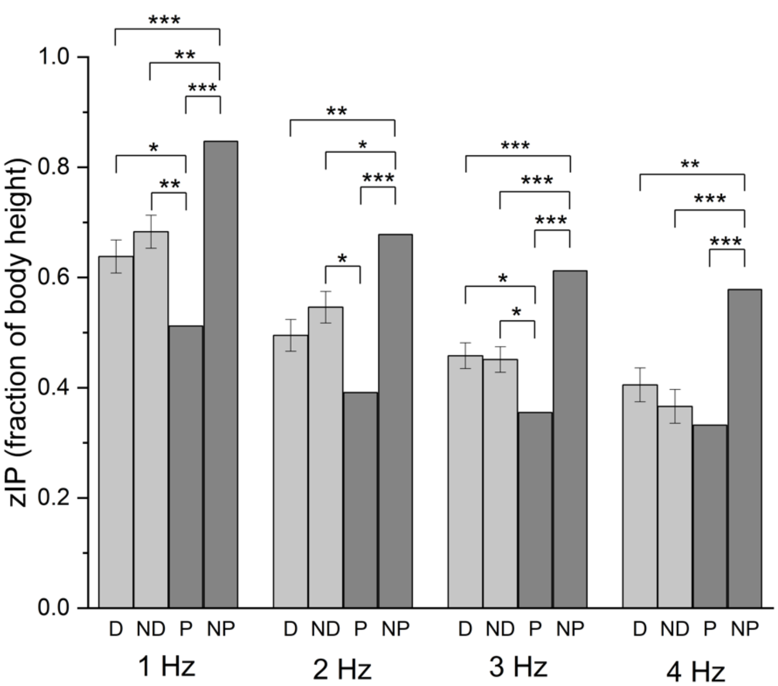

Figure 3.5: At all frequencies the mean zIP values of the dominant and non-dominant legs were similar to each other while the paretic (P) leg was lower than the non-paretic (NP). The paretic leg was typically lower and the non-paretic typically higher than the control participant legs. Error bars indicate ± standard error. The abscissa abbreviations are: D (Dominant control), ND (Non-Dominant control, P (Paretic stroke), NP (Non-Paretic stroke). Ranges of p-values are indicated by * <0.05, ** <0.01, *** <0.001.

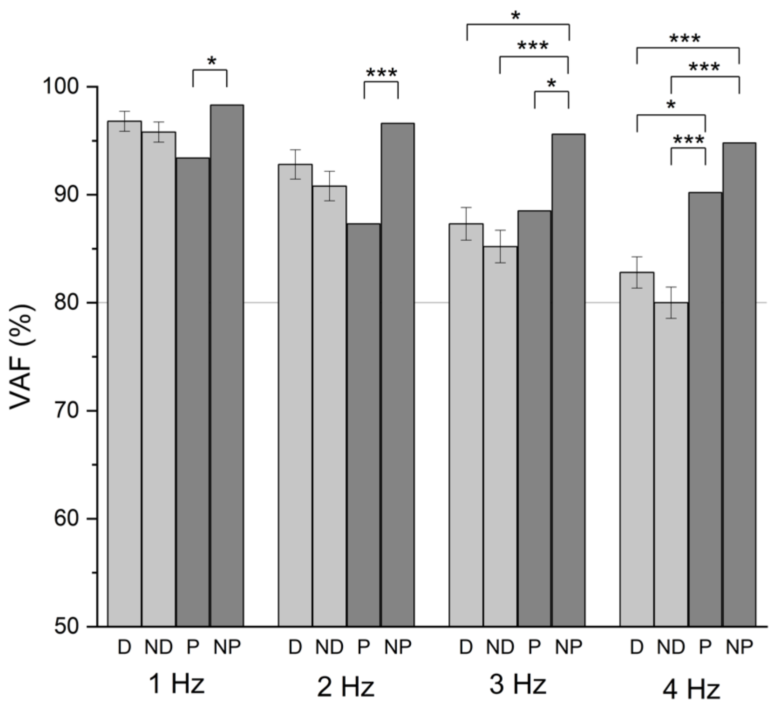

Figure 3.6: The mean VAF (variance-accounted-for) for the relationship between the Fx/Fz versus xCP was above 85% except for the control legs at 4 Hz. The abscissa abbreviations are: D (Dominant control), ND (Non-Dominant control, P (Paretic stroke), NP (Non-Paretic stroke). Ranges of p-values are indicated by * <0.05, ** <0.01, *** <0.001.

The ANOVA models revealed an effect of leg on zIP at all frequencies (df = 3, F>11.9, p≤0.0001). Post-hoc comparisons revealed the following differences (Table 3.1 and Fig. 3.4). At all frequencies, the dominant leg zIP did not differ from the non-dominant leg and the paretic leg zIP was lower than the non-paretic leg. The paretic leg zIP was typically lower than the control legs except at 4 Hz. The non-paretic leg zIP was higher than the dominant and non-dominant legs at all frequencies.

The zIP did not systematically correlate with either mean Fz or with xCP variation (standard deviation of xCP) for the non-paretic leg and both legs of controls (Fig. 3.6). In the paretic leg there was moderate correlation with mean Fz at 2 and 4 Hz and moderate to strong anti-correlation with xCP variation at 1, 2, and 3 Hz (Fig. 3.6).

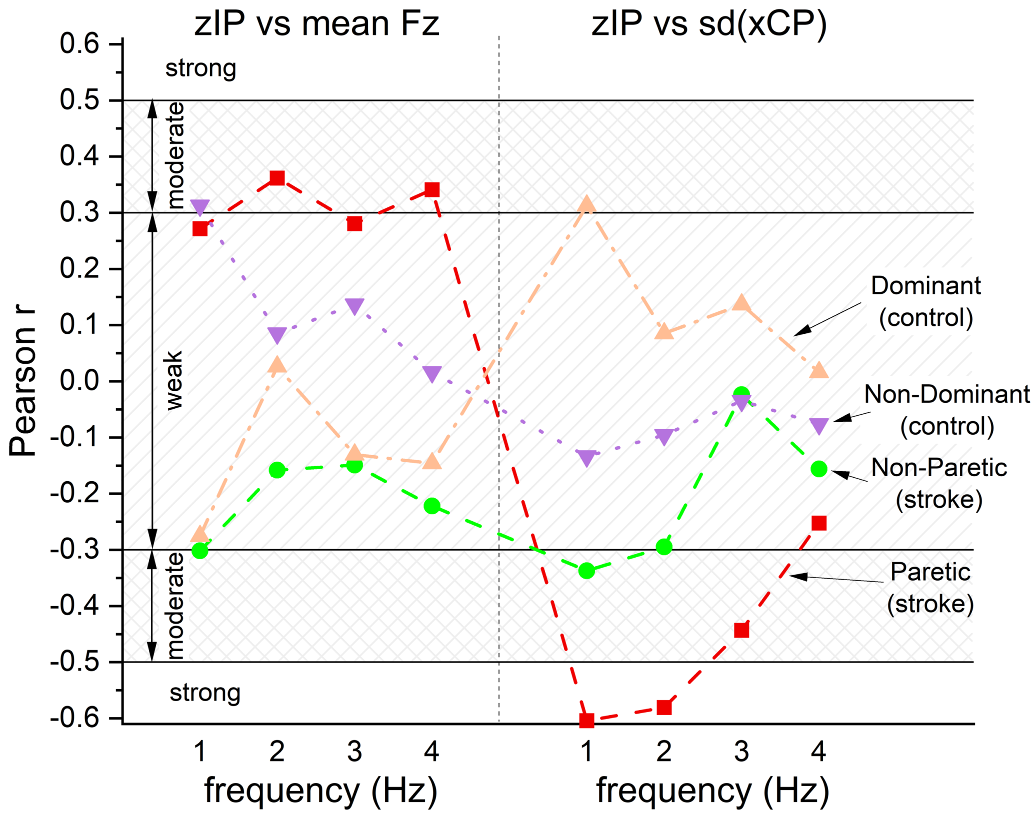

Figure 3.6: zIP was weakly correlated with vertical force on the foot (mean Fz) except for the paretic leg where some moderate correlations were observed. The zIP was weakly correlated with CP variability (sd xCP) except for moderate anti-correlation for the paretic leg at 1–3 Hz.

Discussion

For both groups, IP behavior was observed (high correlation between Fx/Fz and xCP, Fig. 3.2) as indicated by VAF values above 80% for most observations (Fig. 3.5). The lowest consistent VAF occurred for the control group at 4 Hz (Fig. 3.5) which also contained only a relatively small fraction of the overall signal power. The presence of IP behavior in both the paretic and non-paretic legs suggests that the inter-joint coordination necessary to produce an IP has persisted in this cohort despite stroke-induced neurological disruption and/or compensatory adaptations.

The observed zIP dependence on frequency post-stroke demonstrates a distinct systematic control modulation in each of the paretic and non-paretic legs. Across frequencies the zIP values of the paretic leg were generally lower, while those for the non-paretic leg were generally higher when compared to both the control legs (Figs 3.3, 3.4). While these observations do not reveal specific neural mechanisms of the control disruption, we may better understand the meaning and implications of the altered zIP if we consider previous studies of IP behavior.

Earlier work on standing discussed the implications of IP location relative to the CM (Boehm et al., 2019). For small lean angles of the nearly upright human caused primarily by rotation about the ankle, a supra-CM IP assists in standing balance by producing torque about the CM and horizontal force on the body each with the necessary sense to return the body to vertical. These supra-CM IP kinetics are consistent with the kinematic description of standing balance generally observed to cope with minimal standing perturbation called ankle strategy (Creath et al., 2005) where ankle joint motion dominates relative movement between body segments. Mechanical simulation shows that the hip and ankle joint torques that drive a double inverted pendulum model with exclusive ankle motion (hip-to-ankle segment and supra-hip segment move as a single rigid body) are indeed those that produce a supra-CM zIP (Fig. 3.3 labeled ‘rigid’ on plot). The moment of inertia for rigid body motion about the ankle joints is relatively large compared to movements where the torso segment rotates opposite to the hip-to-ankle segment. Thus, the rigid body motion about the ankles responds more slowly to torque inputs, necessitating longer duration inputs characterized by low frequency, and hence the low frequency content is associated with supra-CM zIP.

Having considered the functional utility of the supra-CM zip, lower zIP values will now be presented as an appropriate response to excessive anterior (or posterior) CM excursions. If the CM strays too far anterior or posterior (that is, close to the edge of the base of support), additional horizontal acceleration (in excess of that associated with a supra-CM zIP) may be required to correct CM location. That acceleration can only be produced with horizontal forces from the floor (F, floor on feet), emphasizing the importance of the horizontal forces despite a claim that they ‘play a minor role in human balance’ (Schut et al., 2020). If the body were to remain rigid above the ankle, that force line of action would pass through an IP located just above the CM (Gruben and Boehm, 2012). Producing the increased horizontal F required to accelerate the CM horizontally would require the CP to be located further anterior or posterior. However, foot length presents a limitation to CP excursion, thus limiting horizontal force (under the assumption of rigid body motion). To achieve larger horizontal force within the constraint of foot length, it is necessary to increase the deviation of F from vertical (assuming a relatively constant vertical component of F). Greater deviation in F orientation relative to CP changes results in a lower zIP than the rigid body case. This lower zIP is the product of torque ratios that accelerate the torso and leg segments with opposite rotational sense (Fig. 3.3 Thip/Tankle < 0.25). The lower inertia of this kinematic mode results in a quicker response, requiring more brief inputs that would appear as higher frequency components. Thus, a zIP lower than the rigid case is consistent with the hip joint motion generally observed as a response to large perturbations termed hip strategy (Creath et al., 2005). Again, mechanical simulation demonstrates that the hip and ankle joint torque ratio associated with increased hip angle change are those that produce a zIP lower than the rigid case (Appendix A). In summary, it appears that humans, even after stroke, employ various torque ratios (zIP's) with relative dependence on frequency that correspond, at least in sense, to the varying inertia of the body encountered by the generated movement patterns.

A controller that commands torque ratios *other* than those typically observed (control legs Fig. 3.3) at a given frequency is likely to exhibit some deficiency in retaining upright posture. Both the paretic and non-paretic limbs, while showing similar relative changes in zIP across frequency as seen in control legs, show a distinct shift in zIP at each frequency (stroke legs Fig. 3.3). This could disrupt postural control in the following manner. Consider, as an example, that at about 2 Hz the dominant legs utilize a torque ratio of 0.1 that corresponds to a zIP of 0.5. However, the paretic legs activate that same control (joint torque ratio = 0.1) at about 1 Hz and the non-paretic at 4.5 Hz (Fig. 3.3). The 1 Hz activation by the paretic leg may be too prolonged for the inertia presented by the movement pattern driven by that torque ratio (0.1), leading to excessive motion, while the 4.5 Hz activation of the non-paretic limb may be too brief to cause sufficient motion. Future studies are warranted to evaluate if these hypothesized frequency shifts are experienced as deficiencies and if these abnormal control patterns also present during single leg stance, a critical component of walking and stair climbing, tasks that are more difficult post-stroke.

The standing zIP may be related to F control during a seated lower extremity task. Humans with and without history of stroke, pushing on a pedal while seated, generated F with multi-muscle temporal synchronization (Gruben and López-Ortiz, 2000; Rogers et al., 2004) that is qualitatively similar to the synchronization associated with IP in the present study. Furthermore, the paretic leg relative inter-muscle activation differed from control and non-paretic legs (Rogers et al., 2004) which may stem from the same neurological insult producing the present shifted zIP. While qualitatively similar, direct comparison of these results is impeded because 1) the pedal constrained CP to the ball of the foot and 2) the frequency content of the seated F variations was outside the frequency range evaluated here.

The finding that the paretic leg exhibits a lower zIP could instead be interpreted as a shift of each coordination (zIP) to a different frequency, and vice versa for the non-paretic leg. At present there is insufficient evidence to determine if a frequency shift or zIP shift is the better explanation. Regardless of the mechanism, the practical implication is that the rate at which each coordination is employed will differ from that chosen by non-disabled peers and will thus likely be a disruption to the ability to remain upright.

Decreased mean Fz and decreased variability in xCP were not associated with lower zIP as evidenced by the mostly weak correlation coefficients (Fig. 3.6). It does not appear that the altered zIP is solely a function of weight bearing asymmetry (unequal mean Fz between paretic and non-paretic legs) or reduced CP motion, but rather is due to an altered coordination between the orientation of F and xCP.

A limitation of the present analysis is restriction to the sagittal plane. Due to the distinct differences in body linkage mechanics between the frontal and sagittal planes, control differences may be expected (Deniskina et al., 2001) and future work will be required to determine if an IP also emerges in that plane and/or possibly in three dimensions. Observed inter-joint synergies post-stroke (Sánchez et al., 2017) are not restricted to the sagittal plane and thus could provide insight into frontal plane F patterns captured by this IP approach.

In summary, IP behavior can be viewed as a characterization of the neural strategy to address the combination of whole-body rotational and translational accelerations required to stand. The paretic and non-paretic legs following stroke produce patterns of F variation distinct from each other and from both legs of control participants in a manner that reveals potential balance deficits not described by traditional CP analyses. The remarkably consistent disparity between paretic and non-paretic legs, and the specific deviations from non-disabled participants, suggests that IP analysis reveals a specific disruption in the temporal coordination of multi-joint control that may be a useful tool for further understanding of post-stroke balance deficits and in developing rehabilitation interventions.

Chapter 4: Reliability and Validity of the Force Intersection Point in the Assessment of Human Quiet Standing Balance

published as:

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Abstract

Background: This study evaluated psychometric properties of the Intersection Point Height, derived from ground-on-feet force characteristics, as a tool for assessing balance control. We compare this metric with traditional center of pressure measurements.

Methods: Data from a public dataset of 146 participants, divided into younger (<60) and older (≥60) adults, were analyzed. Clinical tests included Short Falls Efficacy Scale-International, International Physical Activity Questionnaire-Short Form, Trails Making Tests A and B, and Mini-Balance Evaluation Systems Test. Reliability and validity were assessed through the intra-class correlation coefficient (ICC(3,1)) for mean Intersection Point Height in each test condition and Spearman's Rho between summative Intersection Point Height (the sum of intra-condition mean values across all test conditions within one subject) and other variables of interest, respectively.

Findings: Mean Intersection Point Height showed good to excellent reliability (ICC=0.712-0.901), similar to that of CPvel (ICC=0.733-0.922) and greater than that of varCPx (0.475-0.768). Summative Intersection Point Height exhibited strong convergent validity with Trails Making Tests A and B (rho=0.49, p<0.001) and Mini-Balance Evaluation Systems Test (rho=-0.47, p<0.001). At most, a weak to moderate (rho=0.39-

0.49, p<0.001) was found between intra-condition mean Intersection Point Height with center of pressure metrics. Intra-condition mean Intersection Point Height demonstrated weak to moderate convergent validity with several clinical measures (rho=0.32-0.52, p<0.001). In contrast, at most, a weak to moderate (rho=0.39-0.49, p<0.001) association was found between intra-condition mean Intersection Point Height with center of pressure metrics

Interpretation: The Intersection Point Height is a reliable and valid balance measure. Further, we believe that it is a more comprehensive evaluation than center of pressure metrics.

Introduction:

Successful balance control that maintains upright posture is dependent on the complex interaction between sensorimotor systems and the environment (Forbes & Chen, 2018). Balance control impairments increase the risk of falling (Horak, 2006), a leading cause of death in adults age 65 and older, and a leading cause of injury across age groups (*Centers for Disease Control and Prevention, National Center for Injury Prevention and Control. Web-based Injury Statistics Query and Reporting System (WISQARS) [online]*., n.d.). Isolating and defining salient features of balance control is an important step for predicting future falls and developing targeted rehabilitation techniques for high fall-risk populations.

One method of assessing balance control is static posturography, which quantifies center of pressure (CP) movements in quiet standing on a rigid support surface (e.g., CP area, velocity, and displacement; and the spectral frequency distribution of these components) (Visser et al., 2008). Many CP metrics have demonstrated good reliability across adults (Baltich et al., 2014; Demura et al., 2008; Li et al., 2016). Additionally, CP metrics have been shown to have moderate to good convergent validity with the Berg Balance Scale, a commonly used clinical measure of balance and fall risk (Li et al., 2016).

A recent meta-analysis of the ability of the characteristics of CP displacement to differentiate fall risk in older adults concluded that several CP metrics have demonstrated success in distinguishing fallers from non-fallers (Quijoux et al., 2020). Yet, another meta-analysis (Kozinc et al., 2020) concludes that the precise optimal conditions for such measurement (e.g., available sensory input, inclusion of a secondary task, stance position) has yet to be determined. These authors highlight the opportunity for further research to examine relative changes in CP metrics between such task conditions to provide additional information on sensory aspects underlying control. Indeed, Quijoux et al. argue that the neurophysiological determinants of the various center of pressure parameters are not well-defined"

CP metrics have an inherent limitation in describing the neuromotor response to the mechanical demands of balance. One crucial variable in balance control is the ground-on-feet force (F). However, CP metrics fail to capture the orientation of F, which is a key aspect that influences changes in angular and translational momentum of the body (Gruben & Boehm, 2012a). Relative joint torques between body segments determine F orientation, and thus it represents the final common output of the sensorimotor mechanisms that enable balance (Gruben & Boehm, 2012b). CP metrics describe only

F's point of application (the CP), and by omission of F orientation, lack a mechanistic link with control of whole-body balance. It is not surprising then that CP metrics fail to provide a comprehensive assessment of fall risk.

Our lab previously developed a metric which combines F orientation *and* CP to assess control of whole-body motion. When observing the lines of action of F in the sagittal plane during a period of quiet standing, these lines converge at an intersection point (IP). See Appendix B for a more detailed explanation of the IP and its relation to the mechanics of standing. Various balance challenges have been shown to alter the *height* of the intersection point (zIP) (Dutt-Mazumder & Gruben, 2021; Yamagata et al., 2021) and optimal control simulation has replicated zIP behavior in a standing human model (Shiozawa et al., 2021). By capturing the relationship between F orientation and CP, zIP is capable of fully describing F behavior, the sole factor in modifying the body's angular momentum during quiet standing (Gruben & Boehm, 2012a). Accordingly, zIP may provide new perspective on motor control of standing in health and disease such as stroke (Bartloff et al., 2024).

Reliability and validity of zIP have not been established. In this study, we assessed 1) the immediate test-retest reliability of zIP alongside that of common CP metrics, 2) the association between zIP and CP metrics, and 3) the association between zIP and measures of physical activity and balance. We hypothesized that 1) zIP would have moderate to excellent reliability, 2) that zIP would be weakly to moderately correlated with CP metrics, and 3) that zIP would be at least moderately correlated with physical activity and clinical balance assessments.

Methods:

This study analyzed a public posturography dataset with 163 participants (Santos & Duarte, 2016). The database includes demographic information and scores on standardized measures of falls risk, cognition, and physical activity. Participants were excluded from analysis in cases of missing posturography data (n = 1) and those with at least one or more disabling condition(s): hearing and vestibular (n = 8), visual (n = 2), musculoskeletal (n = 3), visual and musculoskeletal (n = 1), hearing and visual (n = 1), and intellectual (n = 1). The included participants were divided into two groups, n = 83 younger (> 18 & < 60 years) and n = 63 older adults (≥ 60 years). We analyzed the Short Falls Efficacy Scale-International (SFES-I) (Kempen et al., 2008), International Physical Activity Questionnaire-Short Form (IPAQ-SF) (Craig et al., 2003), Trail Making Test (TMT) (Bowie & Harvey, 2006), and the Mini-Balance Evaluation Systems Test (mini-BEST) (Franchignoni et al., 2010) (Appendix C). Additionally, we calculated and analyzed CP metrics and zIP (Appendix C).

Posturography:

Each participant’s 60-second duration CP time-series was assessed during three trials in four conditions presented in a condition-randomized order: firm surface with eyes open (Condition 1), firm surface with eyes closed (Condition 2), foam surface with eyes open (Condition 3), and foam surface with eyes closed (Condition 4). Posturography data were collected with a commercial platform (OPT400600-1000; AMTI, Watertown, MA, USA) and amplifier (Optima Signal Conditioner; AMTI, Watertown, MA, USA) using a sampling frequency of 100 Hz (Appendix C).

Posturography data was used to calculate root mean square (RMS) of CP velocity (CPvel), and variance of CP along the x-axis (varCPx. zIP was calculated using previously reported methodology (Boehm et al., 2019) which utilizes spectral decomposition of the force components (Fx, Fz, xCP) into 16 bins of 0.2 Hz width centered at 0.6 Hz through 3.8 Hz. Next, principal component analysis was applied for each bin to extract the relationship of xCP relative to Fx/Fz, which is linear for an exact IP and with the slope yielding IP's height (zIP). Spline fitting to zIP vs frequency data compensated for occasional insufficient power in individual frequency bins. Measured forces were referenced to the surface of the force plate, necessitating accounting for the height of the foam under the feet in zIP calculations (Appendix C). Herein, zIP is reported as fraction of body height.

Statistical Analyses:

All statistical analyses were conducted using R (version 3.6.3 or later) (*RStudio Team. (2021). RStudio: Integrated Development for R. Boston, MA: RStudio, PBC. Retrieved from Https://Www.Rstudio.Com/*, n.d.). Between-group demographic comparisons used two-tailed t-tests or chi-squared tests. A p-value cutoff of 0.001 was applied to determine statistical significance in all analyses. P-values were adjusted using Holm's method during the comparison of zIP and CP data. Between-group differences for the SFES-I, TMT-A and TMT-B, and mini-BEST were assessed with Wilcoxon Rank Sum tests. Chi-squared tests were used to examine between-group differences for the IPAQ-SF.

ZIP analysis metrics were *mean zIP* to summarize zIP across frequency bins for each standing trial, *intra-condition mean zIP* to summarize across all three trials for each participant within each condition, and *summative zIP* to summarize across the four conditions for each participant (Appendix B Fig. B3). Reliability of *mean zIP* across the repeated trials within each condition was assessed with the intraclass correlation coefficient (ICC[3,1]) (Koo & Li, 2016). Using the 95% confidence interval of the ICC estimate as a guide, values less than 0.5 = poor, between 0.5 and 0.75 = moderate, between 0.75 and 0.9 = good, and greater than 0.90 = excellent (Landis & Koch, 1977).

Due to non-normality of the datasets, convergent validity of zIP was evaluated with Spearman's correlations for each condition's *summative zIP* and *intra-condition mean zIP* separately with SFES-I total score, TMT-A and TMT-B times, and mini-BEST total score. Associations between zIP and CP metrics were assessed using Spearman's correlations for *intra-condition mean zIP* with CPvel and varCPx for each condition. P-values for these tests are unadjusted.

Table 4.1
Demographics

Variable	Older (n = 63)	Younger (n = 83)	p-value
Age (years) [a]	71.4 (6.4)	27.5 (7.6)	< 0.001
Sex (female) [b]	51 (81.0%)	53 (63.9%)	0.03
Body Mass Index (kg/m^2) [a]	25.5 (3.1)	22.2 (2.9)	< 0.001
Taking medications [b]	55 (87.3%)	38 (45.8%)	< 0.001
Reported at least 1 fall [b]	15 (23.8%)	20 (24.1%)	0.99
Reported multiple falls [b]	2 (3.2%)	9 (10.8%)	0.15
Number of Falls [a]	0.3 (0.5)	0.7 (2.4)	0.12

[a] = mean (standard deviation) with a p-value from a two-tailed t-test.
[b] = n (%) with a p-value from a chi-square test.

Results:

Data from 146 participants; (83 younger [< 60 years] and 63 older [≥ 60 years] adults) were analyzed (Table 4.1). *Intra-condition mean zIP* values for each condition ($p < 0.001$) and *summative zIP* value ($p < 0.001$) were higher for older adults compared to younger adults (Table 4.2). Older adults had significantly higher mean CPvel ($p < 0.001$) and significantly greater mean varCPx than younger adults for conditions 3 ($p < 0.001$) and 4 ($p < 0.001$), which involved standing on foam.

Table 4.2
Between-group Differences for zIP and CP Metrics

Variable	All	Older (n = 63)	Younger (n = 83)	p-value
zIP Metrics				
Intra-condition mean zIP				
(Condition 1)	0.612 (0.065)	0.651 (0.043)	0.582 (0.063)	< 0.001
(Condition2)	0.618 (0.071)	0.658 (0.047)	0.588 (0.071)	< 0.001
(Condition 3)	0.617 (0.054)	0.655 (0.041)	0.589 (0.044)	< 0.001
(Condition 4)	0.626 (0.051)	0.660 (0.038)	0.601 (0.044)	< 0.001
Summative zIP	2.473 (0.208)	2.623 (0.121)	2.359 (0.187)	< 0.001
varCPx				
(Condition 1)	0.027 (0.029)	0.029 (0.028)	0.026 (0.030)	0.49
(Condition 2)	0.030 (0.030)	0.027 (0.021)	0.032 (0.035)	0.31
(Condition 3)	0.114 (0.054)	0.136 (0.049)	0.098 (0.053)	< 0.001
(Condition 4)	0.132 (0.066)	0.168 (0.076)	0.105 (0.041)	< 0.001

Abbreviations: zIP = height of the intersection point (fraction of body height); CPvel = root mean square of the center of pressure velocity; varCPx = variance of the center of pressure along the anterior-posterior axis. These data are presented as mean (standard deviation) with a p-value from a two-tailed t-test comparing older to younger.

Immediate Test-retest Reliability of the Mean zIP and CP Metrics:

Across all participants, *mean zIP* within each condition had ICCs = 0.809 – 0.901 (Table 4.3). Additionally, ICC values for *mean zIP* within each condition ranged from 0.741 to 0.886 and 0.712 to 0.851 for younger and older participants, respectively. Across all participants, ICCs for CPvel ranged from 0.803 to 0.922 and ICC values were 0.502 to 0.768 for varCPx within each condition. The ICC values for CPvel were 0.866 to 0.917 for younger and 0.733 to 0.829 for older participants. VarCPx had the lowest ICC values, from 0.475 to 0.666 for younger and 0.501 to 0.736 for older participants.

Table 4.3
Reliability for zIP and CP Metrics

Test Conditions	*Mean zIP* ICC(3,1) (95% CI)	CPvel ICC(3,1) (95% CI)	varCPx ICC(3,1) (95% CI)
All Participants (N = 146)			
Condition 1	0.809 (0.759, 0.853)	0.803 (0.760, 0.841)	0.625 (0.556, 0.689)
Condition 2	0.901 (0.872, 0.924)	0.856 (0.823, 0.884)	0.502 (0.422, 0.579)
Condition 3	0.838 (0.794, 0.875)	0.922 (0.903, 0.938)	0.625 (0.556, 0.689)
Condition 4	0.834 (0.789, 0.872)	0.893 (0.868, 0.915)	0.768 (0.719, 0.812)
Younger Participants (n = 83)			
Condition 1	0.763 (0.681, 0.831)	0.866 (0.824, 0.901)	0.606 (0.511, 0.693)
Condition 2	0.886 (0.841, 0.921)	0.890 (0.855, 0.919)	0.475 (0.367, 0.580)
Condition 3	0.741 (0.653, 0.814)	0.917 (0.890, 0.939)	0.658 (0.571, 0.737)
Condition 4	0.784 (0.707, 0.846)	0.877 (0.837, 0.909)	0.666 (0.580, 0.743)
Older Participants (n = 63)			
Condition 1	0.712 (0.603, 0.803)	0.733 (0.647, 0.806)	0.666 (0.566, 0.754)
Condition 2	0.851 (0.785, 0.902)	0.825 (0.763, 0.875)	0.615 (0.507, 0.713)
Condition 3	0.790 (0.702, 0.859)	0.813 (0.748, 0.867)	0.501 (0.378, 0.618)
Condition 4	0.747 (0.647, 0.829)	0.829 (0.768, 0.878)	0.736 (0.650, 0.808)

Abbreviations: ICC = intraclass correlation coefficient; CI = confidence interval; zIP = height of the intersection point; CPvel = root mean square of the center of pressure velocity; varCPx = variance of the center of pressure along the anterior-posterior axis.

Table 4.4
Correlational Analyses

	Spearman Rho (95% CI)	p-value
Associations with Intra-condition Mean zIP		
Condition 1		
Age	0.58 (0.46, 0.68)	< 0.001
SFES-I (Total)	-0.09 (-0.25, 0.07)	0.268
TMT-A (Time)	0.39 (0.24, 0.52)	< 0.001
TMT-B (Time)	0.37 (0.22, 0.5)	< 0.001
Mini-BEST (Total)	-0.36 (-0.49, -0.21)	< 0.001
Condition 2		

Age	0.53 (0.4, 0.64)	< 0.001
SFES-I (Total)	-0.14 (-0.29, 0.03)	0.099
TMT-A (Time)	0.36 (0.21, 0.49)	< 0.001
TMT-B (Time)	0.32 (0.17, 0.46)	< 0.001
Mini-BEST (Total)	-0.32 (-0.46, -0.17)	< 0.001
Condition 3		
Age	0.69 (0.59, 0.76)	< 0.001
SFES-I (Total)	-0.2 (-0.35, -0.04)	0.014
TMT-A (Time)	0.49 (0.35, 0.6)	< 0.001
TMT-B (Time)	0.52 (0.4, 0.63)	< 0.001
Mini-BEST (Total)	-0.46 (-0.58, -0.32)	< 0.001
Condition 4		
Age	0.6 (0.49, 0.7)	< 0.001
SFES-I (Total)	-0.19 (-0.35, -0.03)	0.019
TMT-A (Time)	0.45 (0.31, 0.57)	< 0.001
TMT-B (Time)	0.48 (0.34, 0.59)	< 0.001
Mini-BEST (Total)	-0.49 (-0.6, -0.35)	< 0.001
Associations with Summative zIP		
Age	0.69 (0.59, 0.76)	< 0.001
SFES-I (Total)	-0.18 (-0.33, -0.02)	0.031
TMT-A (Time)	0.49 (0.35, 0.6)	< 0.001
TMT-B (Time)	0.49 (0.36, 0.61)	< 0.001
Mini-BEST (Total)	-0.47 (-0.58, -0.33)	< 0.001
***Association between Intra-condition Mean zIP* and CPvel**		
Condition 1	0.39 (0.24, 0.52)	< 0.001
Condition 2	0.42 (0.28, 0.55)	< 0.001
Condition 3	0.49 (0.36, 0.61)	< 0.001
Condition 4	0.4 (0.26, 0.53)	< 0.001
***Association between Intra-condition Mean zIP* and varCPx**		
Condition 1	0 (-0.17, 0.16)	0.969
Condition 2	-0.05 (-0.21, 0.11)	0.519
Condition 3	0.13 (-0.03, 0.29)	0.109
Condition 4	0.25 (0.09, 0.39)	0.003

Abbreviations: CI = confidence interval; zIP = intersection point height; SFES-I = Short Falls Efficacy Scale - International; TMT =, Trail Making Test; mini-BEST = mini Balance

Evaluation Systems Test; CPvel = root mean square of the CP velocity; varCPx = variance of the center of pressure along the anterior-posterior axis.

Associations Between zIP and Other Metrics:

Associations between *intra-condition mean zIP* values for each condition and other metrics were found to be as follows: TMT-A (Condition 1: rho = 0.39, $p < 0.001$, Condition 2: rho = 0.36, $p < 0.001$, Condition 3: rho = 0.49, $p < 0.001$, Condition 4: rho = 0.45, $p < 0.001$), TMT-B (Condition 1: rho = 0.37, $p < 0.001$, Condition 2: rho = 0.32, $p < 0.001$, Condition 3: rho = 0.52, $p < 0.001$, Condition 4: rho = 0.48, $p < 0.001$), and mini-BEST (Condition 1: rho = -0.36, $p < 0.001$, Condition 2: rho = -0.32, $p < 0.001$, Condition 3: rho = -0.46, $p < 0.001$, Condition 4: rho = -0.49, $p < 0.001$). There was no association between *intra-condition mean zIP* values and SFES-I (Condition 1: rho = -0.09, $p = 0.268$, Condition 2: rho = -0.14, $p = 0.099$, Condition 3: rho = -0.2, $p = 0.014$, Condition 4: rho = -0.19, $p = 0.019$). (Table 4.4)

The associations between *summative zIP* and other metrics found to be as follows: TMT-A (rho = 0.49, $p < 0.001$) and TMT-B (rho = 0.49, $p < 0.001$), as well as with mini-Best (rho = -0.47, $p < 0.001$). There was no association between *summative zIP* and SFES-I (rho = -0.18, $p = 0.031$). (Table 4.4)

For all conditions, *intra-condition mean zIP* was associated with mean CPvel (rho = 0.39-0.49, $p < 0.001$), but it was not associated with varCPx (rho = -0.05-0.25, $p > 0.001$) (Table 4.4). *Intra-condition mean zIP* for each condition and *summative zIP* were associated with age (rho = 0.53-0.69, $p < 0.001$) and rho = 0.69, $p < 0.001$, respectively).

Discussion:

This is the first study to investigate the psychometric properties of zIP. Our data demonstrates that zIP is a reliable and valid measure of balance and that it has comparable or superior reliability as common CP metrics. Along with the greater potential explanatory power of zIP due to the incorporation of both CP location and F orientation, these data support our contention that zIP may provide additional value in quantifying balance ability compared to CP metrics.

Immediate Test-retest Reliability:

Among all conditions for all participants, the reliability of *mean zIP* ranged from good to excellent whereas the reliability of CP metrics ranged from moderate to excellent. *Mean zIP* had moderate to good reliability in younger and older adults. Comparatively, the reliability of CP metrics ranged from poor to excellent in the younger and from moderate to good in the older group.

Our results agree with previous studies that demonstrated the reliability of CP measures. For instance, the reliability of the Sensory Organization Test, which utilizes CP data to assess balance in various sensory conditions, has been established for healthy adults (ICC = 0.9) (Grove et al., 2021; Harro & Garascia, 2019). In healthy adults, CP path length has excellent reliability under conditions of rigid platform eyes open, rigid platform eyes closed, and foam pad eyes open (ICC = 0.93, 0.90, and 0.90, respectively) (Baltich et al., 2014). Establishing the reliability of zIP achieves an important prerequisite for translation of zIP to clinical use, as the clinician can be confident in the precision of zIP when re-testing the same person in the same conditions.

Associations Between zIP and Other Metrics:

Summative zIP demonstrates moderate convergent validity with each of the capacity-based outcome measures (i.e., the ability to perform a given task) including TMT-A, TMT-B, and mini-BEST. *Intra-condition mean zIP* shows similar strength in conditions 3 and 4 (foam surface), but weak validity for conditions 1 and 2 (firm surface).

Summative zIP is similar to the TMT and the mini-BEST in that each is a global measure. Summative zIP comprehensively assesses balance across sensory contexts, while the mini-BEST evaluates multiple aspects of balance, and the TMT captures diverse cognitive functions. In contrast, *intra-condition mean zIP* reflects condition-specific demands. Foam introduces additional postural challenge by altering proprioceptive input and requiring greater motor control. The moderate convergent validity in foam conditions suggests sensitivity to such factors. Together, *summative* and *intra-condition mean zIP* provide a comprehensive assessment of balance as a global metric while also having the capability to capture specific aspects related to challenging conditions.

The associations between zIP and the TMT as well as the mini-BEST support zIP as a potential falls risk metric. The TMT assesses visual-motor processing speed, attention, and executive functioning and the mini-BEST is an assessment of balance and mobility performance. TMT results in combination with a Random Forest Model have been found to be a good predictor of falls in the acute neurologic population (Mateen et al., 2018).

These authors posed that the relationship between TMT performance and falls may stem from the ability of the TMT to measure executive function and processing speed. The mini-BEST involves challenges to cognition (dual-tasking) and attention (obstacle negotiation) and has a demonstrated ability to identify fallers in post-stroke and Parkinson's populations (Franchignoni et al., 2010). These common domains between the mini-BEST and the TMT and the association performance on these tests with zIP suggests that zIP may capture the influence of executive functioning on balance.

Because of the moderate to good convergence of CP metrics in adult populations with established measures of balance such as the Timed Up-and-Go and Berg Balance Scale (Condron et al., 2002), it is important to consider the relationships between intra-condition mean zIP and CP metrics. Examining the 11 comparisons between *intra-condition mean zIP* and CP metrics, six show moderate correlations, one shows a weak correlation, and four show no correlation. The tenuous relationship between zIP and CP metrics supports our contention that *intra-condition mean zIP* and CP metrics measure different aspects of balance and that zIP captures specific aspects of balance that go beyond what traditional CP metrics measure (Condron et al., 2002; Li et al., 2016). The ability of zIP to provide a direct link to the angular motion mechanics required for standing, which CP lacks by itself, further supports our assertion that zIP taps into unique motor control characteristics.

Implications for Further Research and Clinical Translation:

The differences in measurement capability between CP metrics and zIP, along with their similar reliability, justifies further investigation of zIP's utility as a balance metric. By representing not just CP movement but control of F orientation as well, zIP provides potential for greater sensitivity to small changes in balance control. Currently used measures of balance and fall risk are limited by their lack of responsiveness to small differences in ability.

Considering that zIP can be measured with relatively inexpensive equipment and requires 60 seconds (or less) of quiet standing, broad clinical adoption is feasible from a cost and time standpoint. Further examination is needed to determine the strength of zIP as a fall risk assessment. Specifically focusing on its ability to capture and differentiate small differences in motor control ability would provide valuable insights into the responsiveness of zIP as a balance metric and its potential clinical utility.

Table 4.5
Between-group Differences for Self-report and Capacity-based Measures

Variable	Older (n = 63)	Younger (n = 83)	p-value
SFES-I (Total) [a]	9.0 (7.0 - 11.0)	10.0 (8.0 - 12.5)	0.06
IPAQ-SF (Low) [b]	7 (11.1%)	19 (22.9%)	< 0.001
IPAQ-SF (Moderate) [b]	23 (36.5%)	45 (54.2%)	< 0.001
IPAQ-SF (High) [b]	33 (52.4%)	19 (22.9%)	< 0.001
TMT-A (Time) [a]	49.4 (34.6 - 60.5)	19.8 (16.7 - 24.9)	< 0.001
TMT-A (Errors) [a]	0.0 (0.0 - 0.0)	0.0 (0.0 - 0.0)	0.56
TMT-B (Time) [a]	163.0 (91.0 - 238.9)	46.5 (36.7 - 59.0)	< 0.001
TMT-A (Errors) [a]	2.0 (1.0 - 3.0)	0.0 (0.0 - 1.0)	< 0.001
Mini-BEST (Total) [a]	18.0 (17.0 - 21.0)	24.0 (23.0 - 26.0)	< 0.001

Abbreviations: SFES-I = Short Falls Efficacy Scale-International; IPAQ-SF = International Physical Activity Questionnaire - Short Form; TMT = Trail Making Test; mini-BEST = mini Balance Evaluation Systems Test.

[a] = presented as median (inter-quartile range) with a p-value from a Wilcoxon Rand Sum test.
[b] = presented as N (%) with a p-value from a chi-square test.

Limitations:

The current study utilizes a large dataset with multiple accepted measures of balance for study of reliability and validity, including the opportunity to derive any CP metric desired; however, we chose a small representative sample of possible CP metrics based on their high occurrence of utilization across balance studies. When analyzing zIP in trials requiring standing on foam, it was necessary to estimate the amount of foam displacement based on subject mass and foot length to determine zIP (Supplement 4B). While this method was used consistently across subjects, it is a possible source of variability. There were unexpected trends across age in this study population; surprisingly, the older group reported a similar level of fear of falling on SFES-I, higher levels of activity on IPAQ-SF, and fewer falls compared to the younger group (Table 4.5). This is despite significant differences observed between groups on TMT-A, TMT-B, and mini-BEST.

Conclusion:

zIP is a reliable measure of balance with a demonstrated ability to distinguish between healthy younger and older adults. The larger potential for description of motor control of balance contained in zIP versus traditional CP metrics, paired with the finding of comparable or better psychometric properties, suggests that zIP should be considered as an improvement upon traditional measurement of balance in that it may be a more comprehensive measure of motor control. Further psychometric testing is warranted to determine reliability of zIP across longer spans of time, mimicking clinically feasible testing intervals. Additional studies should examine the responsiveness of zIP to small changes in balance ability associated with aging, disease, or response to interventions, further development of zIP as a tool for falls prevention, and the usefulness of zIP to drive referrals for rehabilitation.

Chapter 5: Independent Foot Force Regulation for Postural Stability During Walking: Insights from Force and Speed Perturbations

Manuscript in Progress:

Jennifer Bartloff, David Brown, Negar Moradian, Mansoo Ko, Scott Hetzel, Kreg Gruben

Abstract

Background: Xi, a metric of postural control in the single limb stance phase of walking, emerges as a point in space arising from the systematic relationship between the orientation (θ) and center of pressure (CP) of the ground-on-foot force (F) after accounting for heel-to-toe rollover.

Purpose: To investigate the regulation of Xi in the presence of two gait perturbations: (1) unilateral limb speed changes on a split-belt treadmill and (2) progressively increasing posterior forces (PF) applied at the center of mass during tied-belt and split-belt conditions.

Hypotheses: H1) Xi location would be preserved for the leg experiencing the same speed between tied and split walking, while changing for the other leg, suggesting independent regulation, and **(H2)** Xi location would elevate with increased magnitude of posterior force in both tied- and split-belt walking, suggesting an angular momentum stabilizing response for the perturbation.

Population: 17 non-neurologically impaired adults (25.2 +/-2.51 years)

Protocol: 30 seconds of tied (same speed), then 30 seconds of split ("perturbed" leg (PL) at 50% of tied speed, "control" leg (CL) at 100% of tied speed). Participants repeated this sequence under conditions of differing magnitudes of posteriorly directed resistance force applied at the waist ("PF", 0, 5, 10, 15, and 20% body weight (BW))

Statistics: Linear mixed effects models were used to determine effects of perturbations on each leg for Xi location and peak knee flexion angle (PKA).

Results: Belt condition was found to have an effect on Xi height for the perturbed leg (PL) but not for the control leg (CL). There was an effect of belt condition on Xi horizontal location and PKA for both limbs. The presence of PF resulted in small, yet significant, posterior shifts of Xi and increased PKA for both limbs. However, there was no effect of PF on Xi height.

Conclusion: The uniform effect of posterior force (PF) on Xi horizontal location for both PL and CL, along with a change in Xi vertical location experienced only by the PL in response to a unilateral change in belt speed, supports the concept of independently regulated but coordinated limb control. PKA results match expectations for kinematic changes under the presented perturbations.

Significance: This study builds on previous findings that there exist two independent controllers of legs for walking. Comparing healthy walking Xi strategy data to that of individuals with neurological impairments, such as stroke, will likely reveal differences in this aspect of control.

Introduction

Humans adapt their gait mechanics to overcome diverse challenges, sometimes requiring distinct responses from each stance limb (Hinton et al., 2020; Reisman et al., 2010; Torres-Oviedo et al., 2011). However, the mechanisms that maintain postural stability during single limb stance (SLS) via controlling ground-on-foot forces (F) to sustain the subtasks of support, braking, propulsion, and angular momentum regulation remain poorly understood. During unaided walking, F is the sole force acting on the body that can produce torque about the center of mass (CM). Therefore, both its point of application, or center of pressure (CP), and orientation (θ) are crucial determinants of whole-body angular acceleration, which ultimately influences whether a person maintains balance or tips over. To fully understand the control strategies involved in postural stability, it is essential to consider these elements of F.

Gruben and Boehm (2014) identified a systematic relationship between θF and CP that provides insight into how SLS subtasks interact and postural stability is achieved. Their analysis identified the emergence of a point (Xi) located near the body CM (Gruben & Boehm, 2014). Mathematically, Xi is defined as the intersection point of the lines of action of F when the effects of foot roll-over are removed (Fig. 5.1). This point provides a representation of hip and knee coordination strategies independent of ankle torque, the primary driver of foot roll-over. The position of Xi relative to the CM offers insight into minimization of angular momentum. It was proposed as a regulated target of the motor control system in the maintenance of upright posture.

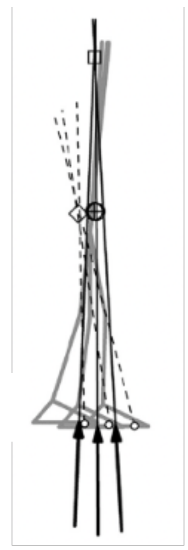


Figure 5.1: Diamond represents Xi, the intersection of lines of action of F with CP relocated to a single point on the foot. Square represents intersection point of F lines prior to CP relocation. Circle represents center of mass.

Given that bipedal walking requires adaptable motor control to meet varying demands and considering evidence from split belt treadmill studies suggesting the existence of independent locomotor networks for each leg during walking (Choi & Bastian, 2007), we theorized that the regulation of Xi occurs independently for each leg. To investigate the regulation of Xi, we explored the effects of two perturbations during treadmill walking: (1) unilateral speed changes on a split-belt treadmill with legs moving backward with matched speed or with one leg moving at half that speed and (2) five levels of posterior force applied to the hips during both tied-belt and split-belt conditions. Our hypotheses were that **(H1)** *Xi location would be preserved for the leg experiencing the same speed between tied and split walking, while changing for the other leg, suggesting independent regulation,* and **(H2)** *Xi location would elevate and shift posteriorly with increased magnitude of posterior force in both tied- and split-belt walking.*

Methods

This study represents a secondary analysis of data collected in an experiment conducted by authors Brown, Moradian, and Ko. Detailed information regarding the study design, participant characteristics, and informed consent can be found in the original publication(Moradian et al., 2023).

Participants were 18 non-neurologically impaired adults (25.2 +/-2.51 years). Due to frequent occurrences of a limb stepping across the center line onto the contralateral belt by one participant, the analyses here included 17 participants.

A force-sensing split-belt treadmill (Bertec Corp., Columbus, OH, USA) with data collection at 2000 Hz and lower body motion capture marker set consisting of 20 markers set collected at 100 Hz were utilized for biomechanical data collection with an eight-Vicon Vantage camera system (Vicon Motion Systems, Inc, Denver, CO, USA).

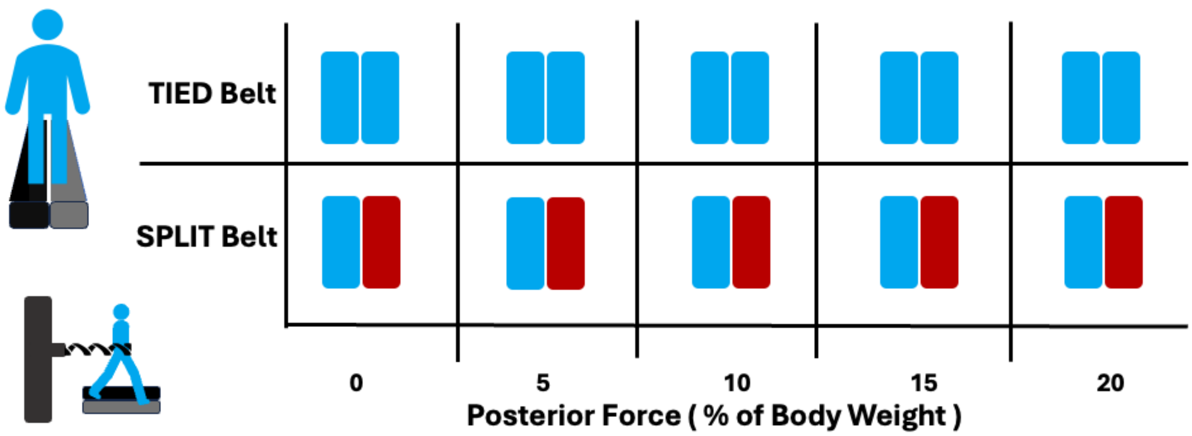


Figure 5.2: Walking protocol. For each participant, trials progressed from tied to split belt conditions over increasing levels of posterior force.

Walking protocol: (See Fig. 5.2) 30 seconds of tied-belt (equal leg speeds), then 30 seconds of split-belt (one leg at 50% speed). For the split-belt condition, the non-dominant leg was selected as 'control leg' (CL) which remained at the same speed as the tied-belt condition, while the dominant leg was the 'perturbed leg' (PL) at 50% speed. Participants repeated this sequence while experiencing five magnitudes of posteriorly directed force on a hip harness denoted 'posterior force' (PF) (0, 5, 10, 15, and 20% body weight (BW)). This force was delivered by a custom apparatus that adjusted for position on the treadmill to keep mean force consistent at the nominal value. The force was not constant but varied through time due to anterior-posterior body motion relative to the treadmill. A slack safety harness prevented injury from loss of balance.

Outcome Measures

Primary Outcome Measures included both the Xi vertical and horizontal location, expressed as a % of body height (BH). Xi was calculated by modulating the measured F to adjust for foot rollover following the methods of Gruben and Boehm (2014) (Gruben & Boehm, 2014). This involves simulated ankle torque modulation to adjustment CP location to the ball of the foot, estimated to be the anterior-posterior location of the 2nd metatarsal head marker. This isolated the coordinated action of knee and hip torque from that of the ankle. CP location was referenced to the center of mass (CM). Per the protocol described previously (Gruben & Boehm, 2012a), adjusting the CP to the ball of the foot (BF) via ankle torque modulation utilized an intersection point height of the knee height (Kvert). The new ratio of horizontal to vertical F after adjustment of CP to BF was calculated as follows:

$$Fhoriz'/Fvert' = (xBF_{COM} + Fhoriz/Fvert * Kvert - xCP_{COM}) / Kvert$$

where Fhoriz is the measured horizontal component of F, Fvert is the measured vertical component of F, xCP_{COM} is the horizontal CP location relative to the CM

For each single limb stance phase in a given trial, Fhoriz'/Fvert' was plotted vs. xCP_{COM} to investigate the linearity of this relationship which represents the F lines-of-action passing near a fixed point in space. A linear regression to this relationship was performed to obtain the slope whose inverse yielded Xi height. Xi horizontal location was equivalent to the x-intercept of that regression.

The secondary outcome measure for this study was peak knee flexion angle during swing phase (PKA). This value was calculated from motion capture data, using lateral thigh, and knee marker. Thigh (A) and shank vectors (B) were defined as vectors from the knee marker to the respective segment marker. The knee joint angle was calculated as the angle between these vectors using the following formula:

$$\theta = \arccos\{A \text{ dot } B / (|A||B|)\}$$

where A is the thigh vector and B is the shank vector.

Statistical Analysis

Analyses utilized R for statistical computing (version 2023.03.0+386, R Core Team, 2021) and were conducted at a 5% significance level. Linear mixed effects models (LME) and generalized linear mixed effects models with binomial family (GLME) from the “lme4” library (Bates et al., 2015), in conjunction with the “emmeans” library (Lenth, Russell V., 2021), were used to estimate means and percentages across interventions and posterior force by including belt condition (2 levels), posterior force (continuous), and their interaction (belt:PF) as fixed effects predictors and subject as a random effect. To assess possible differences between belt condition and PF, LME models with belt condition and PF as fixed effect predictors and subject as a random effect were fit, and post-hoc comparisons were conducted, when applicable, with Tukey’s family-wise adjustment for multiple tests. For the comparison of baseline (tied-belt, 0% PF) differences between dominant (D) and non-dominant (ND) limbs, an additional random effect of paired consecutive left-right gait cycles was included to enhance detection of inter-limb differences rather than stride-to-stride variations.

Results

Inter-Limb Difference for Baseline Condition

In the baseline condition (tied-belt, 0% PF), the Xi was higher for the D limb compared to the ND limb (0.9% BH, p<0.0001). Xi horizontal location did not differ between D and ND limbs (p=0.9). The two limbs did not differ in knee angle (p=0.07).

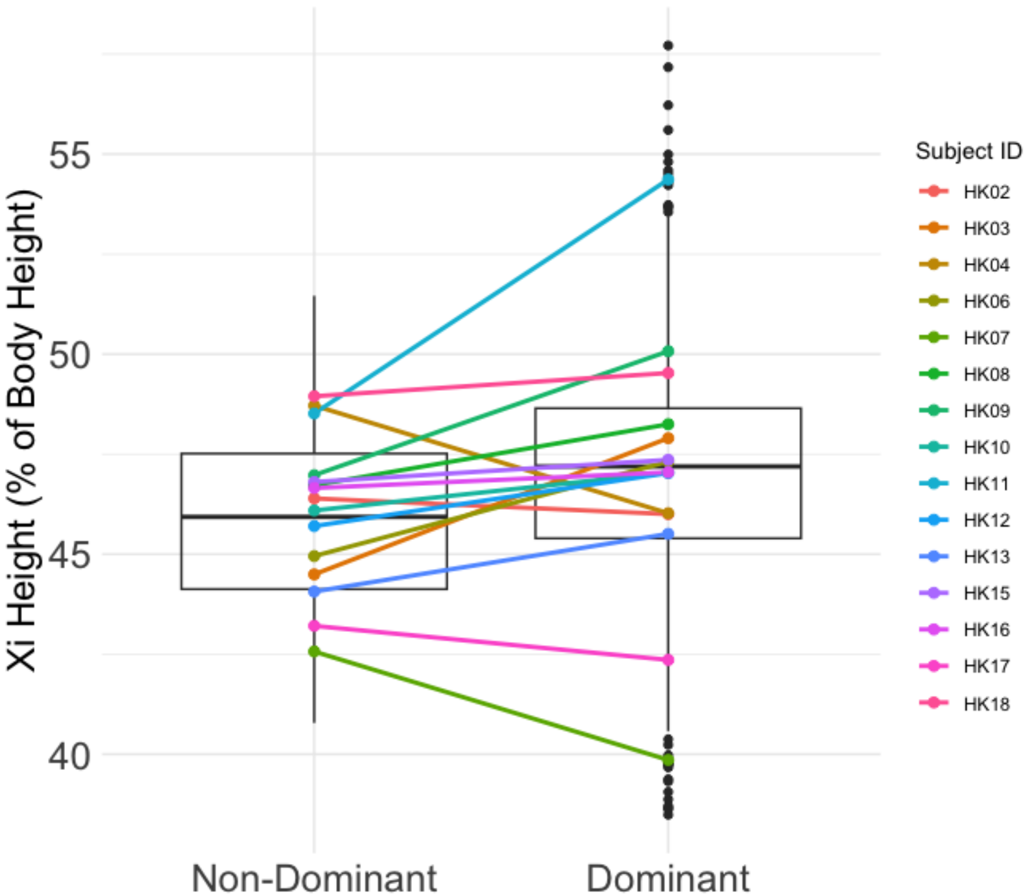


Figure 5.3: Baseline condition (tied-belt, 0% PF) Xi height for dominant and non-dominant limbs. Lines connect limb average for each participant.

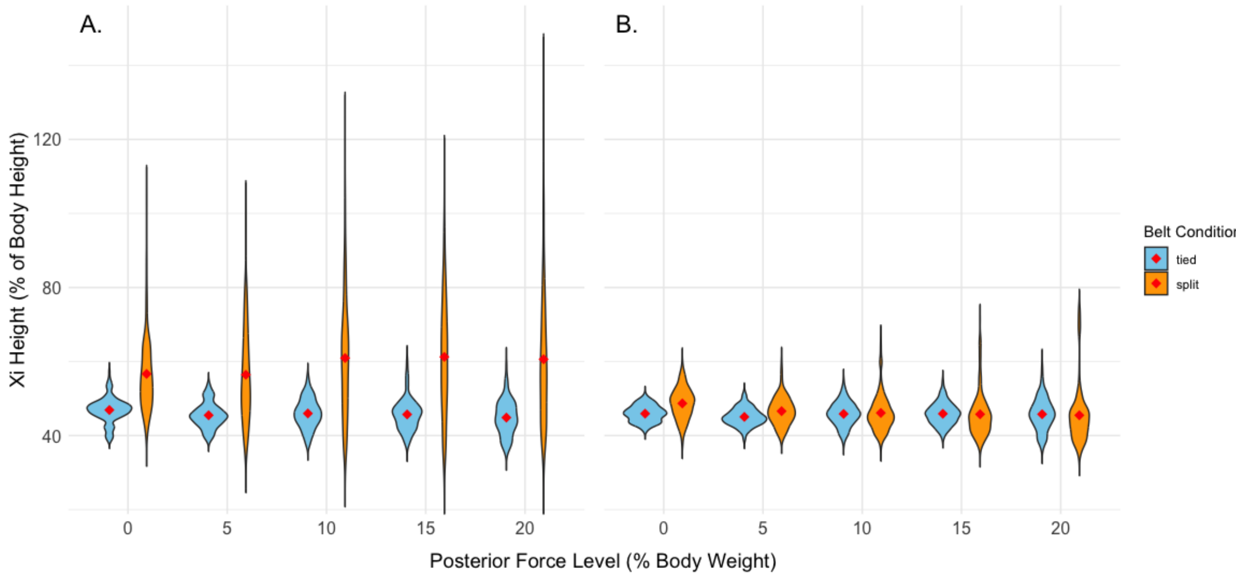


Figure 5.4: A. Xi height for PL; **B.** Xi height for CL; belt condition data represented for each PF level.

Effect of Belt Condition (Tied vs. Split)

Xi height: Xi height in the PL was 12.6% BH higher when that limb was at 50% speed (split-belt). (PL: $p<0.0001$, Control: $p=0.1$).

Xi horizontal location: For the Perturbed limb, the split condition is estimated to result in Xi location of 5.2% of body height posterior to that of the tied condition ($p<0.0001$). For the Control limb, the split condition is estimated to result in a Xi location of 1.4% body height anterior to that of the tied condition ($p<0.0001$).

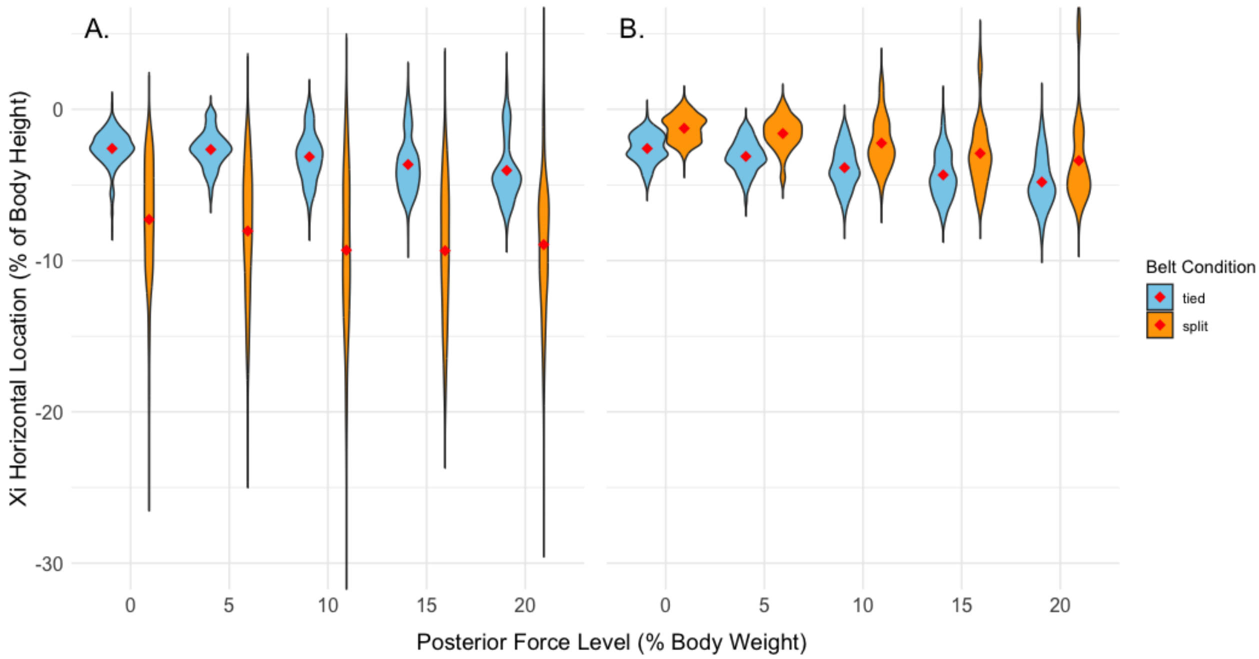


Figure 5.5: A. Xi horizontal location for PL; **B.** Xi horizontal location for CL; belt condition data represented for each PF level.

Peak Knee Flexion Angle: The tied condition consistently resulted in greater knee flexion than the split condition for both limbs (PL: 4.4°, p=0.02; CL: 3.2°, p=0.003).

Effect of Posterior Force (PF)

For both the perturbed and control legs, there was no effect of PF on Xi height (PL: p=0.5; CL: p=0.1), yet the PF did cause a posterior shift in Xi horizontal location (PL: 0.08% BH per 1% BW PF, p=0.02; Control: 0.11% BH per 1% BW PF, p<0.0001).

The PF increased knee angle for both limbs (PL: 0.3 deg/1%BW PF p=0.009, CL: 0.28 deg/1% BW PF p=0.004).

Discussion

Inter-Limb Differences in Baseline Condition:

That an effect of limb dominance is seen on Xi height is not surprising. Leg dominance has been associated with differing neural activation characteristics on fMRI (Kapreli et al., 2006). Given that leg dominance is associated with such differences, it follows that motor control may be executed differently between limbs and thus limb dominance may have an effect on motor control outcome measures. This difference is quite small, 0.9% of body height, and the Xi horizontal location was not impacted by limb dominance.

This differential effect on vertical and horizontal locations of Xi is consistent with current understanding of Xi mechanics. In the baseline condition, without a PF applied to the hips the Xi horizontal location should be consistently close to the CM so that the average torque about the CM is small. A bias in Xi anterior or posterior from the CM will result in a non-zero average torque about the CM, increasing momentum through time and likely leading to a fall. However, increased Xi height does not alter the average torque but instead increases the range of the positive and negative torques about the CM enhancing the righting torque (toward upright) and promoting upright balance. The presence of a higher Xi location in the dominant limb may reflect a tendency of greater reliance on that limb for control of angular momentum.

The presence of a PF requires adjustments in neuromuscular activity to sustain upright posture so that the PF does not accelerate the person posteriorly. An increase in anteriorly directed force of the ground on the feet (F) is needed to avoid posterior translational acceleration of the CM. However, this creates a force couple with the PF that will induce a posterior pitching angular acceleration and lead to a fall. To prevent this, F must be adjusted to produce a counter-torque (anterior pitching). A horizontal shift in Xi itself does not necessarily provide the mechanical solution to this problem because Xi describes only a component of the ground on foot force, the component resulting from removal of foot 'roll-over' mechanics. Foot 'roll-over' mechanics (CP shifting with respect to the foot) also alter F orientation and thus the horizontal component of F. To address the need for anterior pitching torque, the system has multiple options, including shifting the CP posteriorly and modulating F magnitude through the gait cycle. The analysis performed here does account for such responses of the controller as Xi is independent of F magnitude and by definition does not reflect CP location independent of F direction. To fully understand the role of horizontal Xi shifts in the control of whole body angular momentum, a deeper analysis that includes ankle mechanics is required. Such approaches exist and should be included in further study (Gruben & Boehm, 2012b, 2014). Thus, these results exclusively show that the Xi component of the control is altered by PF and do not describe how the system compensates for the mechanical disruption of the PF.

The observation that Xi height was higher in the dominant limb while peak knee flexion angle was the same in both limbs, may be a reflection of the complexity of motor control. While dominance plays a role in more complex or high-demand activities, such as postural adjustments under challenging conditions or sport-specific maneuvers such as kicking or cutting, basic walking biomechanics in the absence of perturbation do not require inter-limb kinematic differences Further, it is possible that certain biomechanical requirements of the walking task are more sensitive to the influence of limb dominance due to the level of fine motor control required. Stance phase in healthy adults is likely to require more fine control than gross motor swing phase movements such as joint position.

Effect of Belt Condition (Tied vs. Split)

The higher and more posterior Xi for the perturbed limb when it was on the slow belt shows that this limb adapted force control in response to the slowed speed. The control limb changed Xi horizontal location only, but this change was much smaller than that of the perturbed limb (Perturbed = 5.2% posterior shift, Control = 1.4% anterior shift). The relatively small anterior shift in the control limb could be a counterbalance to the perturbed limb's adaptation to the speed challenge.

It is well established that under split-belt walking conditions, the limb on the slower belt will adapt to provide an increase in the propulsive component of F (Hagen et al., 2024; Hsiao et al., 2015). This may be due to changes in muscle coordination patterns or just the posteriorly biased posture adopted by that leg. As noted above, a horizontal shift in Xi could alter horizontal F, but will not necessarily do so because of the other mechanisms such as CP and F magnitude that are also available. While the speed change in a split-belt paradigm overtly impacted Xi of the leg on the slowed belt interlimb coordination is also affected by the task asymmetry.

Small effects of belt condition on knee angle were observed for both limbs, with the split condition showing decreased max knee flexion relative to the tied (PL = 4.4°, Control = 3.2°). Previous research has shown that changes in walking speed influence knee joint kinematics, with faster speeds resulting in greater knee flexion. Kirtley et al. showed an increase of 8.6° max knee flexion for every 1 m/s increase in walking speed (Kirtley et al., 1985). It is challenging to make a direct comparison between this study and Kirtley's findings, as this study treated speed as a two-level factor of belt condition (tied vs. split) rather than testing speed in a manner suitable for linear regression analysis. However, the increase in peak knee flexion with increased walking speed is a well-documented phenomenon, and our result for the perturbed limb is consistent with those findings.

Interestingly, belt condition effect on max knee flexion angle was observed not just for the perturbed, but also for the control limb, which did not experience a speed change. This suggests that the influence of belt condition on knee flexion angle extends beyond immediate mechanical effects on the limb subject to speed perturbations. The control limb swing phase kinematics may be influenced by the perturbed limb's elongated stance phase. Perhaps the contralateral increase in stance time influences the control limb swing phase kinematics to adjust for a slower walking speed, resulting in less knee flexion in the split than tied condition. We would then need to justify why the perturbed limb would demonstrate reduced max knee flexion during the split belt condition, when the contralateral (control limb) will experience a relatively shorter stance phase. Riesman et. Al. (2005) cites possible reasons for restoration of kinematic symmetry during split-belt gait trial, including a greater efficiency of symmetric walking patterns vs.

asymmetric and neural interlimb coupling mechanisms. Further, they state the expectation that sensory feedback from both limbs is necessary for adaptation of interlimb coordination (Reisman et al., 2005). The unexpected phenomenon of decreased swing phase max knee flexion angle in the control limb in the split condition may result from complex interactions between these factors.

Effect of Posterior Force (PF)

As discussed previously, the implications of a horizontal shift in Xi location are not fully clear. PF's effect on only the horizontal, and not vertical, component of Xi, likely reflects some importance of this horizontal location. Both the PL and CL demonstrated this shift, though a deeper analysis that incorporates ankle mechanics is necessary to fully understand this result.

The lack of PF effect on Xi vertical location however, reflects an aspect of the Xi control strategy that is clearly consistent with task mechanics. A change to vertical Xi location, which alters torque magnitude range without biasing the mean, is not required by the imposition of a posterior force. Thus, a lack of effect on this aspect of Xi is not surprising. Our original hypothesis had proposed that Xi would raise in both limbs with the presence of PF. That height increase for Xi has been proposed as a means to strengthen the righting torque and thus could be used by the system to compensate for perturbation.

Small but significant effects of PF on knee angle were seen for both limbs, approximately 0.3° for each unit (%BW) increase in PF for each. The presence of this

PF changes task mechanics in a manner that is similar to external forces applied to the body when walking uphill. Just as walking uphill requires greater effort to move against gravity than level-ground walking, the PF mimics this experience, imposing a demand to increase muscle activation and knee flexion during swing phase. The significant effect of PF on both limbs reflects expected alterations in control of kinematics in response to this perturbation.

Limitations

The 30-second walking trials limited measurement of outcomes to the early phase of adaptation in the face of perturbations. With longer exposure times, individuals would likely have displayed a change in their neuromuscular responses, possibly extinguishing the independence of the Xi height response in the split belt condition to reveal a new combined strategy. For the purposes of this analysis, this early adaptation window is desirable to allow this specific window into control strategy early adaptation. Further, the early adaptation phase is interesting in that it holds ecological validity with typical unilateral gait speed perturbations in the real world. Stepping around an obstacle may for instance require a unilateral speed change, but only briefly.

Conclusions

This study builds on previous findings that suggest there exist two independent controllers of leg muscles for walking. Perturbations impacting the whole body (posterior forces applied at the center of mass) affected both legs similarly and with similar magnitudes of response in horizontal Xi location. Perturbations impacting legs differently (tied vs. split belt, where only the dominant limb experienced a change in walking speed) resulted in independent Xi alterations to match individual task needs of each limb. Both limbs responded to the change in belt condition (tied vs. split) for horizontal location of Xi, yet only the perturbed limb experienced a rise in Xi height. Compared to application of the posterior force, the split belt condition imposes a perturbation that is biased toward the perturbed limb while also impacting inter-limb coordination. This was reflected in the Xi strategy in that both limbs were affected, but the perturbed limb to a greater extent.

Future work

Comparing healthy walking Xi strategy data to that of individuals with neurological impairments, such as stroke, will likely reveal differences in this aspect of control. These insights could contribute to the development of novel therapeutic strategies that target the specific muscle coordination associated with the Xi phenomenon, specifically in the context of stroke rehabilitation. Additional work is needed to describe the effect of these task perturbations on the other components of walking, namely the pattern of CP and F direction not captured by Xi.

Chapter 6: Assessing Magnitude and Location of Minimum Foot Clearance Across Interventions in Multiple Sclerosis with Foot Drop

Manuscript in Progress:

Jennifer Bartloff, K. Heidi Fehr, Yisen Wang, Katherine Konieczka, Julia Mastej, Evan Cohen, Scott Hetzel, Peter Adamczyk

Introduction

Multiple Sclerosis, or MS, is an auto-immune disease of the central nervous system that leads to demyelination of axons and impaired neural transmission, resulting in a spectrum of sensorimotor problems unique to each affected individual (Ghasemi et al., 2017). (Ghasemi et al., 2017). A common manifestation of gait dysfunction in MS is foot drop, characterized by difficulty in actively dorsiflexing the ankle. The presence of foot drop often leads to compromised foot clearance, increasing the risk of falls (Cameron & Wagner, 2011). The presence of foot drop not only contributes to higher energy cost of ambulation but also necessitates adoption of compensatory gait behaviors such as hip hiking, circumduction, steppage, and vaulting (Bulley et al., 2015; Knarr et al., 2013). Additionally, persons with MS (PwMS) suffer a high prevalence of motor fatigability, an objective trait defined by a reduction in muscle function and reduced work capacity over the course of a sustained bout of exercise (Enoka et al., 2021). The interplay between foot drop and fatigability often necessitates exaggerated biomechanical compensations during gait, further elevating fatigue and risk of falls.

In response to these challenges, orthotic solutions are commonly prescribed to reduce the toe-down deviation in ankle posture through swing phase, with goals of reducing the incidence of foot scuffing against the ground, mitigating gait compensations, improving walking economy, and reducing likelihood of falls. Two commonly prescribed categories of orthosis include the carbon-fiber ankle-foot orthosis (AFO) and the functional electronic stimulator (FES), a form of neuro-orthosis. Carbon-fiber AFO’s hold the foot at approximately a 90-degree angle relative to the tibia through the gait cycle, are lightweight, and may provide some energy return at pushoff. FES devices, such as the Bioness L300 Go or WalkAide, stimulate the common peroneal nerve to induce tibialis anterior contraction and subsequent dorsiflexion, with initiation timed by a built-in movement sensor at the start of swing phase. The fundamental differences in these mechanisms - passive support from carbon-fiber AFOs vs. active muscle stimulation by FES - results in markedly different effects on the control of the toe-down ankle posture and the demand on muscles. These biomechanical differences may influence the risk of falls and experience of fatigue during gait.

The choice of which style of device to recommend is based on clinician experience, patient preference, and research evidence, the three pillars of evidence-based practice in healthcare. While there is a growing body of research comparing the effectiveness of foot drop interventions in MS and other clinical populations, a comparison of biomechanical implications is typically lacking. Frequent outcome measures in such studies include metabolic cost of transport (COT) of such devices, total distance walked in the six-minute walk test (6MWT) (ATS Committee on Proficiency Standards for Clinical Pulmonary Function Laboratories, 2002; Goldman et al., 2008), 10 meter walk test (Bohannon, 1997), Dynamic Gait Index (DGI) (Shumway-Cook & Woollacott, 2007) or the MS Walking Scale (MSWS), a subjective measure of confidence in walking (Andreopoulou et al., 2018). While these metrics appropriately address the effects of foot drop interventions on gait-related efficiency, endurance, dynamic balance, and confidence, they do not describe how the interventions themselves alter the mechanics of gait, particularly in the presence of motor fatigue, and thus are missing a key descriptor of the effectiveness of these interventions.

A crucial aspect often overlooked is the impact of foot drop interventions on foot clearance, or height of the foot above the walking surface during swing phase, especially in the context of fatigue. One study examining biomechanics of gait in the MS population investigated the importance of minimum toe clearance (mTC), measured as the smallest vertical distance between the walking surface and the toe's motion-capture marker during swing phase. mTC was compared between groups of fallers, non-fallers, and healthy controls, revealing a trend wherein fallers exhibited the lowest foot clearance, while healthy controls the highest (Peebles et al., 2017). Another study (Byju et al., 2016) found a higher probability of tripping with reduced mTC. Taken together, these studies strongly suggest that if the goal is to reduce tripping and falls, it is important to measure clearance. However, mTC, as customarily measured, captures only the height changes of a single marker at the toe – it fails to capture the clearance of the entire foot, which may be important in cases of pathologic gait in which kinematics deviate from normal. In particular, swing phase frontal plane motion, which is likely to vary between the interventions being examined in the present study and in persons with neurologically-rooted gait impairments (Vachranukunkiet & Esquenazi, 2013), necessitates a more comprehensive approach. Minimum foot clearance (mFC) is an enhanced variation on clearance measurements that measures the shortest vertical distance between any part of the foot and the walking surface during swing phase (Fehr et al., 2024; Schulz, 2011). By including all points on the bottom of the foot, mFC can capture a more complete view of foot clearance in gait, enhancing understanding of the impact of foot drop interventions and fatigue on gait biomechanics with particular emphasis on foot clearance as an important marker of trip and fall risk.

A limited body of previous studies observing the impact of fatigue on joint kinematics (Sehle et al., 2011) and typical spatio-temporal gait parameters (Feys et al., 2013; McLoughlin et al., 2016) have been investigated in the context of MS-related fatigue. However, to our knowledge, no studies have investigated changes in mFC with different foot drop interventions, nor any interactions between mFC and the biological trait of fatigability. This study's purpose was to compare mFC and fatigability during gait among three cases: no intervention (NI), use of a carbon fiber ankle-foot orthosis (cfAFO), and use of a functional electrical stimulator (FES). The goal was to provide a more comprehensive evaluation of these interventions in addressing the compounded challenges of foot drop and fatigue in gait among PwMS. Enabled by our lab's novel method of estimating 3D whole foot movement (Fehr et al., 2024; Schulz, 2017), this greater comprehensiveness included an analysis of the location where mFC occurs on the foot. This aim of this regional (forefoot/hindfoot, medial/lateral) analysis was to leverage readily available technology to provide a unique perspective into intervention effectiveness.

Methods

Procedure

Participants in the study were 11 persons with Multiple Sclerosis (PwMS) who exhibited foot drop (FD) as determined by their physician (age 32–65 years, median 45.6; Self EDSS range 2–6, median = 4). Participants provided written informed consent according to procedures approved by the University of Wisconsin-Madison Health Sciences Institutional Review Board. Detailed demographic and baseline characteristics of participants are presented in Table 1. We secured one inertial measurement unit (IMU) sensor (APDM Opal) on each shoe in a pouch affixed to the instep atop the shoelaces and held in place with zip ties. We inserted a low-profile locating fixture with the IMU, and then 3D-scanned participants' feet/shoes (Occipital Structure Sensor) to generate a geometric model of the foot and shoe surface. Then we removed the locating fixture and resealed the pouch while the IMU remained in place. The IMU data were used to reconstruct movement of the foot and estimate mFC in each stride, as detailed below.

Crossover Design

In their first of three visits to the laboratory, participants first performed a 6–minute walk test (6MWT) with no FD intervention (NI). They then received the cfAFO or FES device for an extended device familiarization period of at least one week, followed by a return to the laboratory for another 3D scan and 6MWT with the device. Finally, they received the other device (FES or cfAFO) and repeated the familiarization period and in-lab follow-up test. See Fig. 6.1 for visual overview of study design

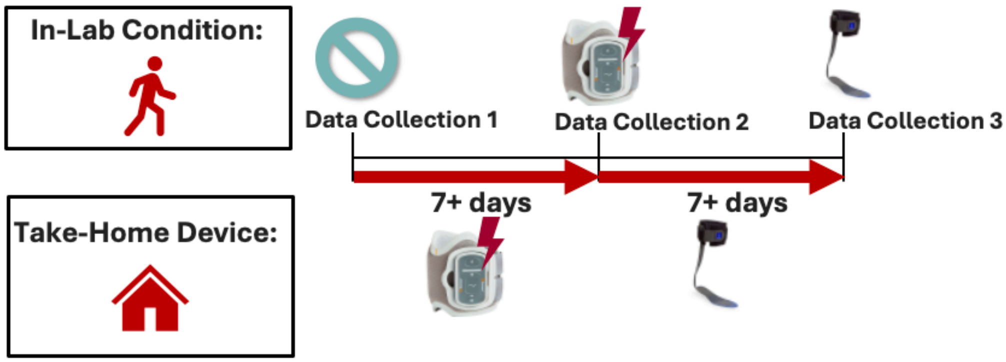


Figure 6.1: Diagram of study crossover

Primary Outcome Measure

The primary outcome measure was minimum foot clearance (mFC) as measured by our novel method of pairing 3D scans with IMU trajectory. The IMU trajectory was reconstructed using an integration-based Pedestrian Dead Reckoning (PDR) algorithm (Wang et al., 2024). This trajectory reconstruction was then combined with points extracted from the 3D scan of the participant's shoe per Fehr et al. (2024) (Fehr et al., 2024) to obtain the global position of each point on the shoe during the 6MWT. The instantaneous foot clearance (IFC) was then determined as the lowest point on the shoe at each instant. We defined an interval of measurement centered on gait events that are readily identifiable with the IMU reconstruction, termed 'forward swing': the period beginning at the instant when the heel is at its maximum vertical position and concluding when the toe achieves its maximum vertical position. Despite the presence of foot drop in our population, this forward swing period was consistently identifiable during straight-path gait.

Average foot clearance (AFC) was recorded for each forward swing as the average of IFCs over every time step (approximately 40 time steps per forward swing phase), (figure 6.2, dashed orange line).

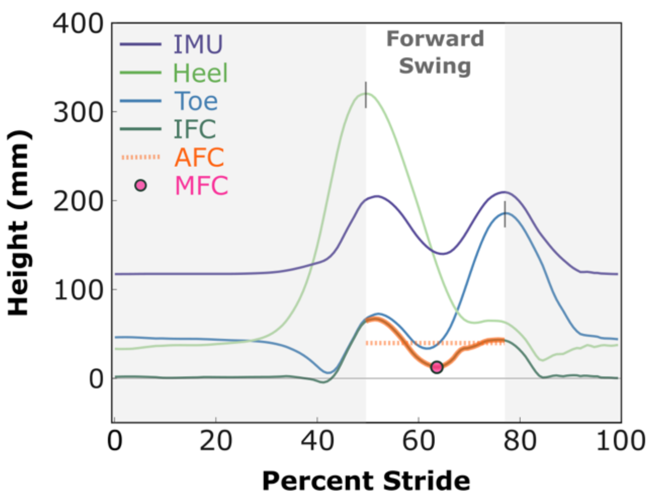


Figure 6.2. Heights of different points on the foot during a single stride.

Minimum foot clearance (mFC) was determined by identifying the lowest IFC during each forward swing period (figure 2, magenta dot). Note that all foot clearance data for these analyses was collected exclusively during the forward swing phase, as this is the portion of gait when a scuff or trip is most hazardous as the foot is moving at its maximum forward velocity (Winter, 1992). This approach was used to exclude data points immediately following toe-off and just before heel strike, periods when the foot is just leaving and entering contact with the walking surface around normal footfalls.

mFC was assessed during each minute of the 6MWT using three methods: 1) Instantaneous Foot Clearance (IFC) for every timestep in a given forward swing (approximately 50 forward swing phases per minute) 2) AFC, and 3) mFC of each forward swing phase. The first two methods provide insight into foot clearance behavior throughout forward swing, while the second offers a more precise perspective on the minimum clearance magnitude that participants typically exhibited during a specific intervention and minute across the 6MWT.

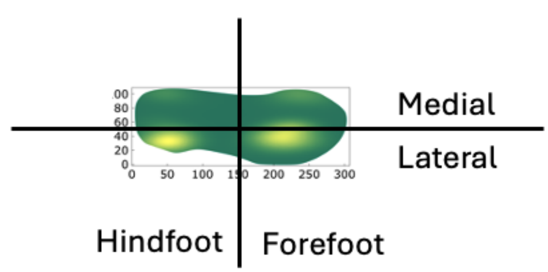
Secondary Outcome Measures

Fatigability: Fatigability was quantified by the Distance Walked Index (DWI): percent change in distance walked in the 6th vs. 1st minute of the 6MWT (Leone et al., 2016). From AFC and mFC, additional measures were computed to further objectively assess fatigability. AFCΔ and mFCΔ, were defined as the difference in AFC or mFC between

Subject	Age (years)	Gender	Self-EDSS Score	Foot Drop Side	6MWT Distance (m)	DWI (%)
1	50	Female	4	R	346.2	-5.8
2	51	Male	2	R	472.7	7.1
3	64	Female	4	L	409.9	1.2
4	33	Female	4	R	237.3	-35.8
5	43	Male	4	R	518.6	-0.2
6	39	Female	2	R	590.5	-7.6
8	45	Female	6	R	271.7	-10.7
9	56	Female	4	L	460.1	-14.9
10	32	Female	2	R	484.2	2.3
11	46	Male	4	L	212.9	-39.4
12	65	Male	4	L	335.9	6.6

Table 6.1: Baseline characteristics of study participants

6MWT minutes 1 and 6. To compute AFC-Index, AFCΔ was divided by the value of AFC from minute 1 and multiplied by 100. This calculation makes AFC-Index



comparable to the DWI in assessing changes over time.

mFC Region: Another outcome metric was the location of mFC on the bottom of the foot. To summarize the distribution of mFC locations, the shoe was divided into four regions by segmenting Medial vs. Lateral and Forefoot vs. Rearfoot regions (See Fig. 6.3). Schulz (2017) demonstrated that minimum foot clearance occurs during the forward swing phase, and that a toe-down position is typical for the early to middle portion of this time window (Schulz, 2017).

In our attempt to characterize effectiveness of interventions for foot-drop, we restricted analysis of mFC in forefoot/hindfoot regions to only the portion of forward swing where a toe-down ankle posture is NOT expected. The forward swing phase of the involved limb was identified and divided into three equal time periods. Based on Schulz (2017), we restricted this regional analysis to the third period of forward swing.

In contrast, analysis of mFC location in Medial/Lateral regions is reported for the entire forward swing phase. According to Neumann (2016), typical frontal plane ankle positioning in the frontal plane during the swing phase is neutral (Neumann, 2016). Consequently, there is no justification for restricting forward swing to a specific time period when assessing the efficacy of FES vs cfAFO in facilitating appropriate frontal plane kinematics. At study's end, participants indicated their preference for which condition they liked best during the week-long take-home familiarization periods: NI, cfAFO, or FES.

Statistical Analysis

Linear mixed effects models (LME) and generalized linear mixed effects models with binomial family (GLME) from the "lme4" (Bates et al., 2015) library, in conjunction with the "emmeans" (Lenth, Russell V., 2021) library of functions, were used to estimate means and percentages across interventions and time points by including intervention (3 levels), time (6 levels), and their interaction as fixed effects predictors and subject as a random effect. To assess for differences between interventions, LME and GLME models with intervention and time as fixed effect predictors and subject as a random effect were fit, ANOVA F-test on the intervention effect while controlling for time was made, and post-hoc comparisons were conducted, when applicable, with Tukey's family-wise adjustment for multiple tests. Analyses utilized R for statistical computing version 2023.03.0+386 (R Core Team, 2021) and were conducted at a 5% significance level.

To visualize the probability of a given part of the shoe serving as the minimum point continuously throughout forward swing phase, heat maps showing the smoothed distribution of mFC locations across all forward swing phases in a bout of walking were created using Gaussian kernel functions.

Coefficient of variation was calculated to examine consistency of timing of the mFC during forward swing.

We interpreted 6-Minute Walk Test (6MWT) results in relation to established criteria for Minimal Detectable Change (MDC) and the Minimal Clinically Important Difference (MCID) to assess for clinical significance of observed changes in walking performance.

The MDC represents the smallest amount of change in the 6MWT that can be detected beyond measurement error and variability, thus providing a benchmark for statistically significant changes. The MCID, on the other hand, represents the smallest change in the 6MWT that is perceived as beneficial by patients or clinicians, reflecting a threshold for clinically meaningful improvement. Including these metrics enhanced the interpretation of intervention effects on functional mobility in the study population.

Results

Average Foot Clearance (AFC), 6MWT

In this study, both cfAFO and FES interventions significantly increased the AFC during the 6MWT compared to NI, with p-values of < 0.0001 for both (Fig. 4). Additionally, cfAFO significantly increased AFC during the 6MWT when compared to FES ($p < 0.0001$).

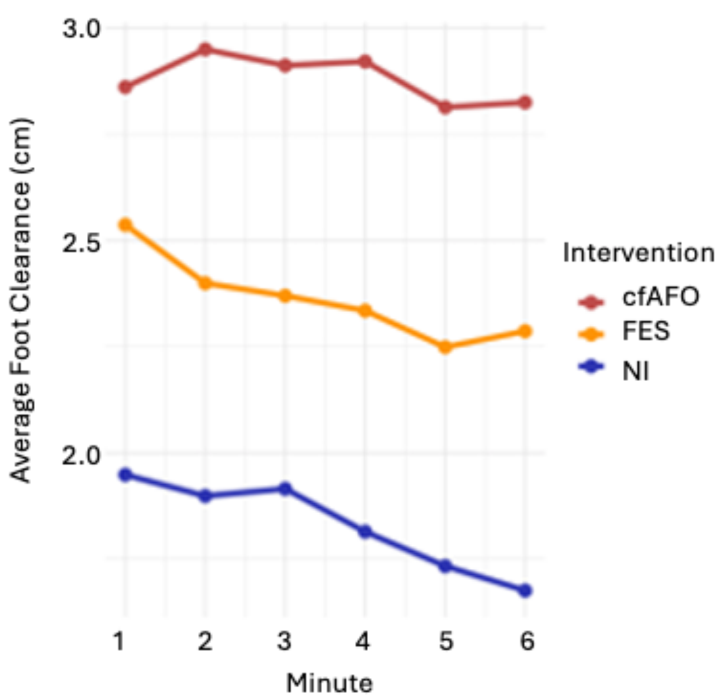


Figure 6.4: Averaged across participants, AFC by minute and intervention

Minimum Foot Clearance(mFC) for each Forward Swing Phase, 6MWT

When taking the mFC for each forward swing phase, a significant increase was observed for each intervention compared to NI. The cfAFO intervention (mean+/-sd) resulted in a mean increase in minimum foot clearance of 0.53 cm (95% CI; $p<0.0001$) compared to the no intervention (NI) condition (mean+/-sd). Similarly, the FES intervention (mean+/-sd) increased the mean minimum foot clearance by 0.24 cm (95% CI; $p<0.0001$) compared to NI. Additionally, the cfAFO intervention resulted in a significant increase in mean mFC compared to the FES intervention by 0.29 cm (95% CI; $p<0.0001$).

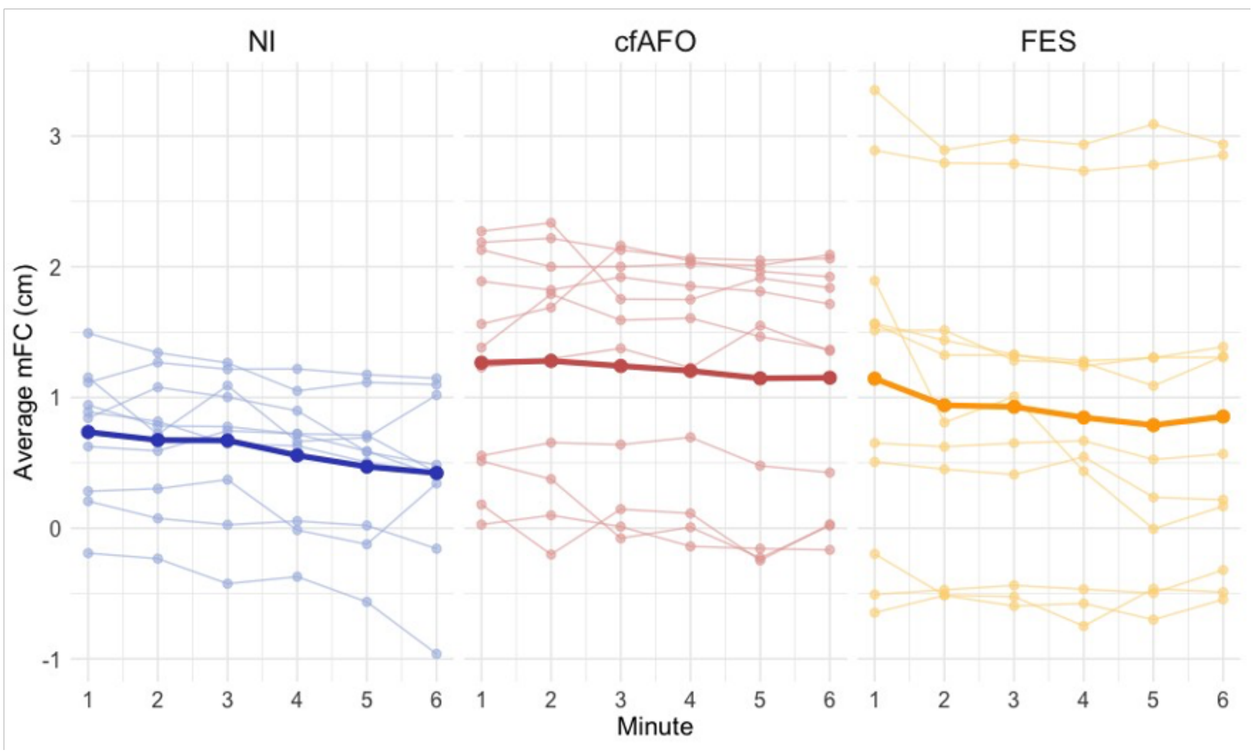


Figure 6.5: In bold: mFC averaged across participants by minute; lighter lines represent individual subject data with mFC averaged by minute for each intervention.

6MWT Total Distance

The cfAFO significantly increased total distance walked compared to NI (p=0.0338) while the FES did not (p=0.0851). The difference in total distance between cfAFO and FES was not significant (p=0.9276).

Subject	NoInt	Sprystep	Bioness
1	346.2	416.7+	412.8+
2	472.7	802.6*	795.9*
3	409.9	396.4	444.3
4	237.3	245.7	320.9+
5	518.6	563.4	531.2
6	590.5	618.3	629.6
8	271.7	287.8	282.7
9	460.1	461.9	423.8
10	484.2	513.5*	508.7
11	212.9	310.6	181.0
12	335.9	384.2	372.2
Mean (95% CI)	**394.5(308.3-480.8)**	**454.6 (321.1-588.2)**	**445.7 (312.2-579.3)**

Table 6.2: 6MWT Total Distances (meters) by Subject and Intervention. Key: *: MDC threshold met vs. NoInt, +: MCID threshold met vs. NoInt.

Distance Walked Index

There was no significant intervention effect on DWI. (p = 0.91 for FES to NI, p = 0.83 for cfAFO to NI. P = 0.58 for FES to cfAFO).

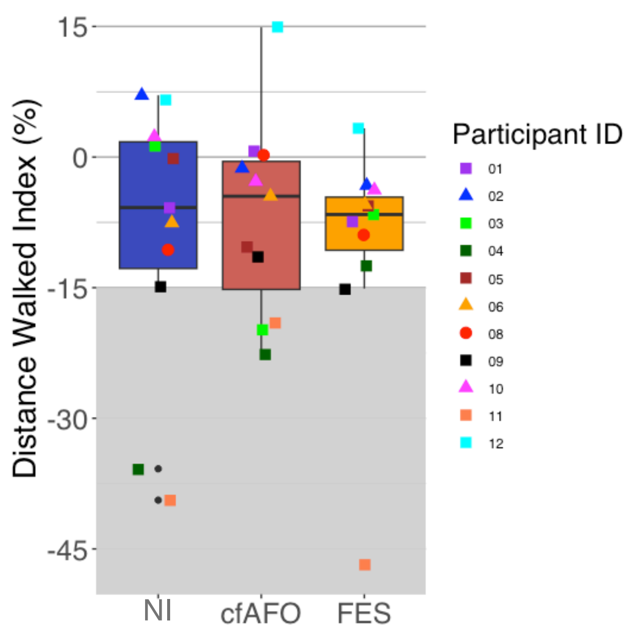


Figure 6.6: Distance Walked Index for each participant by intervention. Gray plot area represents cutoff score for fatigability (-15% and lower). EDSS scores represented by different shapes: triangle = 2, square = 4, circle = 6.

Average Foot Clearance Change (AFCΔ) and AFC-Index

For AFCΔ, there was no effect of intervention. (FES vs. NI: $p=0.9$; cfAFO vs. NI: $p=0.1$).

Additionally, there was no significant effect of intervention on AFC-Index (FES vs. NI: $p=0.4$; cfAFO vs. NI: $p=0.06$

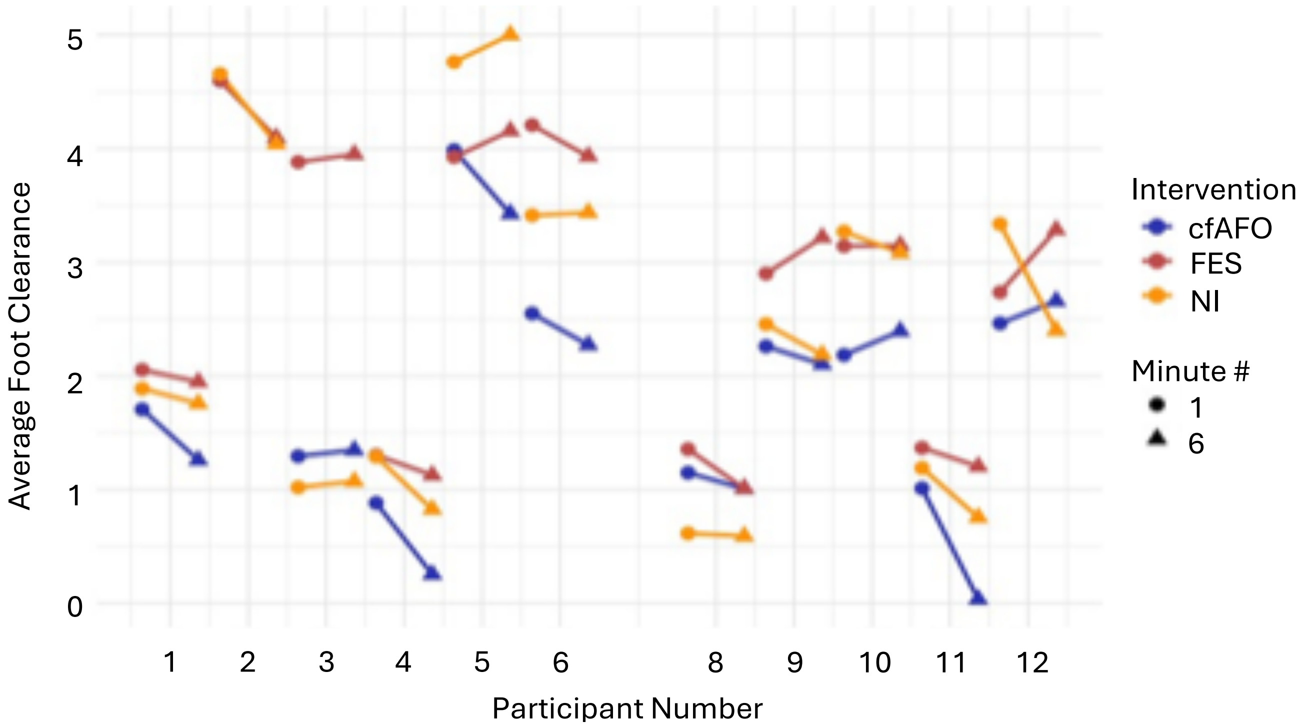


Figure 6.7: Participant AFC from minute 1 (circle) to minute 6 (triangle) by intervention. Most participants' AFC declines, showing a possible effect of fatigue

Minimum Foot Clearance Change (mFCΔ)

There was no effect of intervention on mFCΔ (FES vs. NI: $p=0.8$; cfAFO vs. NI $p=0.2$)

Regional Analysis of mFC Location: Instantaneous Foot Clearance (IFC) throughout Forward Swing

Forefoot:

The first-minute percentage of mFC in the Forefoot region varied across treatments ($p<0.05$), with NI having the highest rate (see Table 6.4 and Fig. 6.8). However, across all interventions over the course of the 6MWT, the percentage of measurements in Forefoot increased. This was most pronounced for NI with an increase from minute 1 to 6 of 14.5 percentage points and less so with cfAFO (2.5 percentage points) and FES (8.2 percentage points).

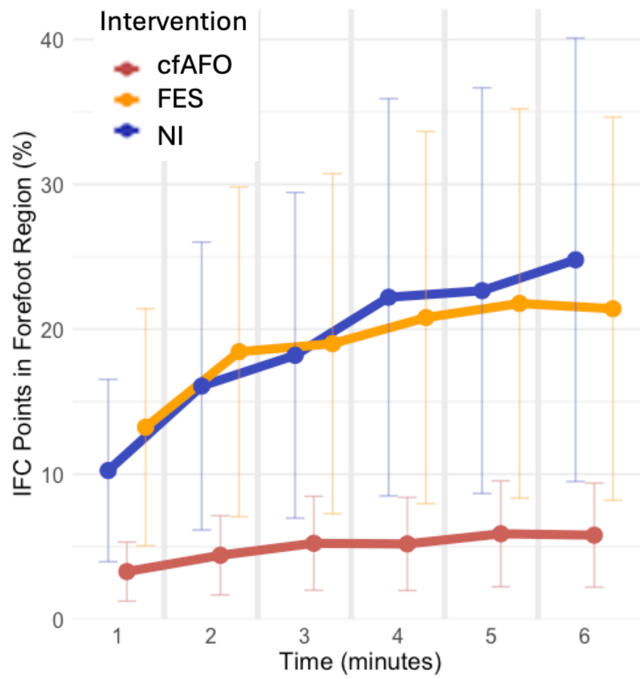


Figure 6.8: IFC points from complete forward swing phases in forefoot region, by minute and intervention. Each datapoint represents average over all participants.

Medial:

In minute 1, the percentage of Medial mFC location was lower in NI than in either intervention ($p < 0.001$). There were significant differences among all three conditions for each minute, with FES consistently displaying the highest percentage and NI the lowest (See Table 6.5 and Figure 6.9).

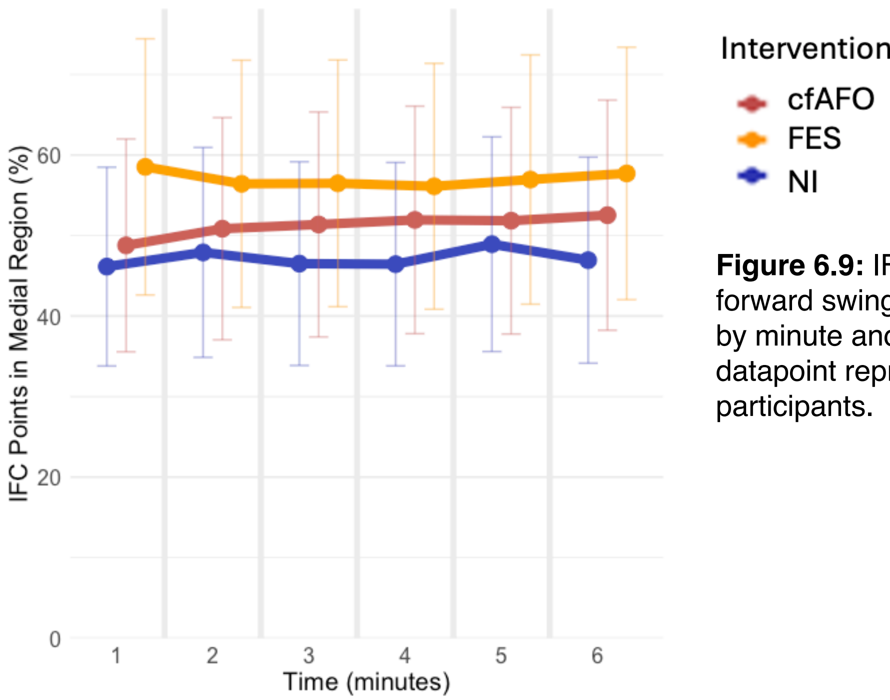


Figure 6.9: IFC points from complete forward swing phases in medial region, by minute and intervention. Each datapoint represents average over all participants.

Minute	NI	FES	cfAFO	Significance
1	46.1% (40.7-51.6)	58.5% (53.0-63.8)	48.8% (43.2-54.4)	A,B,C
2	47.9% (42.3-53.5)	56.4% (50.9-61.8)	50.8% (45.3-56.4)	A,B,C
3	46.5% (41.0-52.1)	56.5% (50.9-61.9)	51.4% (45.8-56.9)	A,B,C
4	46.4% (40.9-52.0)	56.1% (50.5-61.5)	51.9% (46.3-57.5)	A,B,C
5	48.9% (43.3-54.5)	56.9% (51.4-62.3)	51.8% (46.2-57.4)	A,B,C
6	46.9% (41.4-52.5)	57.7% (52.1-63.1)	52.5% (46.9-58.1)	A,B,C

Table 6.5: Probability of mFC Occurrence in Medial Region by Treatment and Minute
Key: A: No Intervention vs FES; B: No Intervention vs AFO; C: FES vs AFO
*capital letters indicate $p < 0.001$; lower case letters indicate $p < 0.05$.

Regional Analysis: minimum Foot Clearance (mFC) Occurrence in Forefoot Region

For all conditions, the probability of the mFC point in each Forward Swing occurring in the Forefoot, rather than Rearfoot region, is high (see table 6, Fig. 9). Both interventions were found to significantly lower the probability of the single lowest point occurring at the forefoot vs. NI ($p < 0.0001$ for both). The comparison between cfAFO and FES was not significant (p=0.9).

Effect of Intervention on Timing of mFC within Forward Swing

Both interventions had a significant impact on the timing of mFC occurrence within forward swing ($p < 1e\text{-}04$) with the cfAFO moving the mFC occurrence to 5.3 percentage points later and FES moving mFC to 4.7 percentage points later, compared to NI. With forward swing lasting just over 0.3 seconds, such changes represent 15-16 milliseconds (ms). There was no between-intervention difference (p=0.4686).

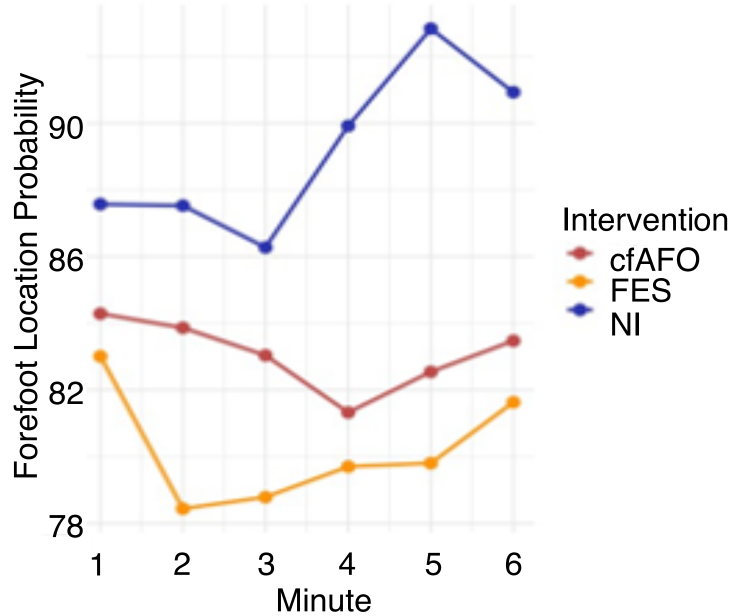


Figure 6.10: IFC points from complete forward swing phases in medial region, by minute and intervention. Each datapoint represents average over all participants

*Effect of Intervention on **Consistency** of Timing of mFC*

There was a significant effect of cfAFO vs. NI on the CV of timing of lowest point in forward swing, with cfAFO having a lower CV (p=0.0001) (see table 6.7). FES also significantly lowered the CV compared to NI (p=0.02). The difference between the interventions for this metric was not significant (p=0.2)

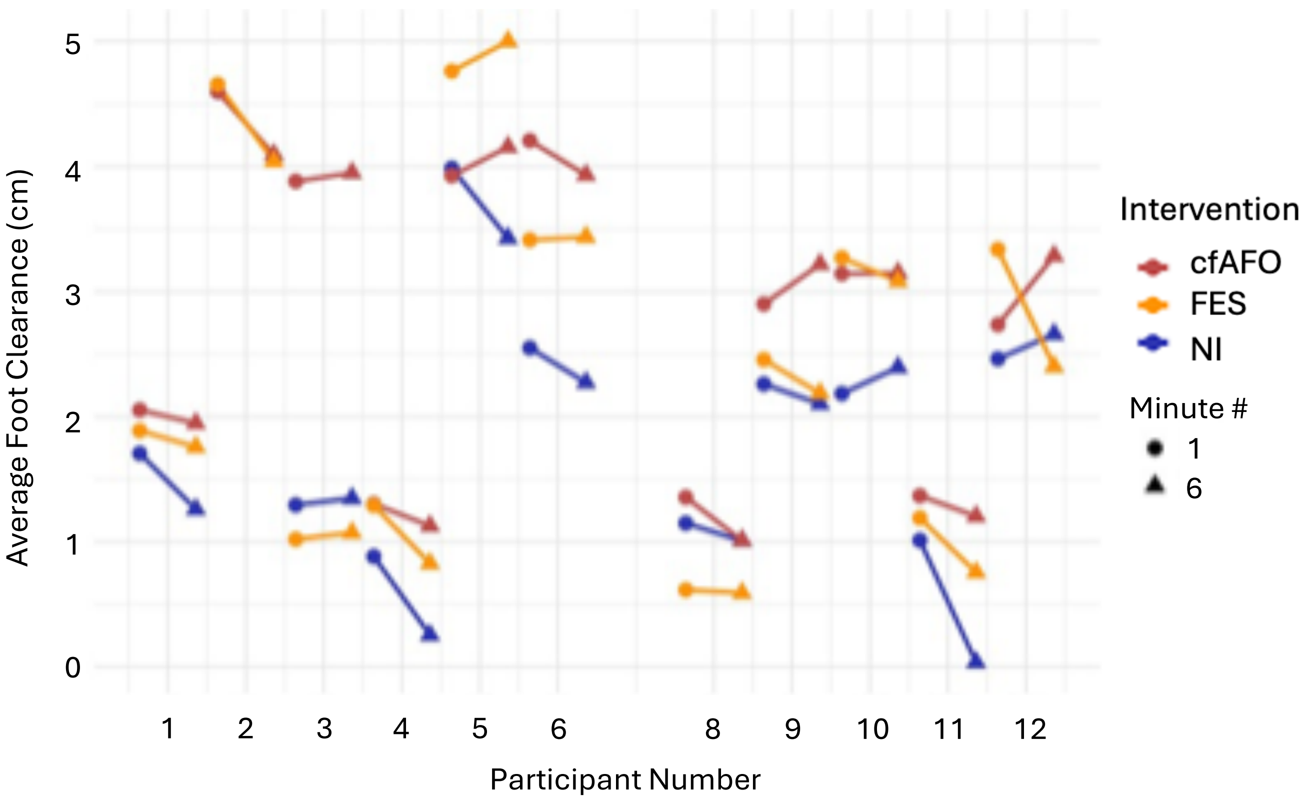


Figure 6.11: Timing of mFC in forward swing stated at % of fwd swing duration. Each point represents the average across participant and minute for a given intervention.

Intervention	Mean	Median	Min	Max
NI	34.1%	22.3%	7.4%	139%
cfAFO	15.8%	10.5%	4.7%	43.1%
FES	27%	14.8%	5.7%	78%

Table 6.6: Coefficient of Variation by Intervention for Timing of Single mFC in Forward Swing

Participant Preference

At the end of the study, participants were surveyed on their preference if they were to pursue a device: FES, cfAFO, or none. 10 of 11 participants preferred a device over NI. Different subsets of 6 of 11 participants' preferences correlated with devices that provided the greatest 6MWT distance, the highest average minimum foot clearance (mFC) during the entire forward swing of the 6MWT, the highest AFC, or the most favorable mFCΔ.

Participant	Preference	Most Favorable DWI	Highest AC-mFC	Highest A-mFC (single point per Forward Swing)	Most Favorable mFCΔ (AC-mFC)	Lowest %mFC in Forefoot	Greatest 6MWT Distance
1	FES	cfAFO	cfAFO	*NI*	FES	cfAFO	cfAFO
2	cfAFO	*NI*	cfAFO	FES	FES	cfAFO	cfAFO
3	cfAFO	*NI*	cfAFO	cfAFO	FES	cfAFO	FES
4	FES	FES	cfAFO	FES	cfAFO	cfAFO	FES
5	cfAFO	*NI*	FES	FES	cfAFO	FES	cfAFO
6	cfAFO	cfAFO	cfAFO	cfAFO	cfAFO	FES	FES
8	FES	cfAFO	cfAFO	*NI*	FES	cfAFO	cfAFO
9	cfAFO	cfAFO	cfAFO	cfAFO	cfAFO	*NI*	cfAFO
10	FES	*NI*	cfAFO	cfAFO	*NI*	cfAFO	cfAFO
11	cfAFO	cfAFO	cfAFO	cfAFO	cfAFO	cfAFO	cfAFO
12	cfAFO	cfAFO	cfAFO	cfAFO	cfAFO	cfAFO	cfAFO

Table 6.7: Participant preferences and best intervention across selected outcome measures. Key: Outcomes highlighted in yellow represent matches with FES, in red represent matches with cfAFO

Discussion

This study investigated the effects of two types of interventions, a carbon-fiber ankle foot orthosis and a functional electrical stimulator, on foot clearance and fatigability during a 6 minute walk test in persons with Multiple Sclerosis and foot drop. Our results highlight the complexity of addressing foot drop while considering fatigability in this population.

Minimum Foot Clearance (mFC) and Average Foot Clearance (AFC)

The significant increase in both AFC and mFC for both interventions compared to NI highlights the effectiveness of these devices in improving foot clearance. This finding is consistent with previous research indicating that both cfAFO and FES can effectively mitigate the risk of tripping and falls in PwMS by enhancing toe clearance during the swing phase of gait. The cfAFO provided a significantly higher AFC and mFC than FES. This suggests that both interventions both improved overall foot clearance and provided consistent increases in the lowest clearance point during forward swing, likely providing a lesser likelihood of scuffing and trips than the NI condition.

6MWT Total Distance (6MWT)

Only the cfAFO significantly increased total distance during the 6MWT – by an average of 60.1 meters – but it did not provide a significant advantage over the FES, which on average improved distance walked by 51.2 meters. Very few participants achieved a change between conditions of NI, FES, and cfAFO that met MCID or MDC criteria. Participants 1 and 2 met thresholds for MCID and MCD, respectively, for both FES and cfAFO vs NI. Participant 4 met MCID for FES vs NI and met MDC for FES vs cfAFO. The superiority of FES in participant 4 matched that individual's device preference at the end of the study. Participant 10 met MDC for cfAFO vs NI, and 11 for cfAFO vs FES. These did not match device preference in either case. Interestingly, a few participants demonstrated shorter walk distances with a device vs. NI. Participants 9 and 11 experienced a reduction in 6MWT distance in FES, and participant 3 experienced a reduction in 6MWT distance with cfAFO. For participant 3, this result was inconsistent with their device preference of cfAFO.

Distance Walked Index (DWI)

On average, the study observed some amount of fatigability across all conditions, as indicated by negative values for the average DWI for each condition (Table 2). Though none of these condition-specific DWI averages met the previously established cutoff for severe fatigability (Leone ref) at -15%, it can be observed on Figure 7 that several participants did meet that criterion for some or all of their conditions. It is interesting to note that participant 9 performed at < -15% at baseline and with FES, but their DWI improved well above the severe fatigability range for the cfAFO condition. Similarly, participant 4 improved to a DWI out of the severe fatigability range with the FES, but not with cfAFO.

Foot Clearance Change and Index Outcomes

Though there was no intervention effect on mFCΔ, mFC-Index, AFCΔ, or AFC-Index, Figure 6 reveals an overwhelming tendency for a reduction in foot clearance from minute 1 to minute 6 across interventions and participants. There are many exceptions without a consistent pattern across the group. This analysis reveals the variability of participant response to each intervention in terms of both AFC and the possible effect of fatigability viewed through the perspective of changes in clearance. At the individual level, participant 11 presented with the worst DWI scores of the group, with NI and FES yielding particularly large declines in walking distance from minute 1 to 6. That participant had a correspondingly poor performance on their AFCΔ for both of these conditions. Participant 4’s NI scores for DWI and AFCΔ were similarly poor, while their cfAFO DWI did meet the -15% cutoff, yet they appear to have a relatively small AFCΔ for cfAFO. No consistent trends between DWI and AFCΔ are apparent.

Regional Analysis of IFC Location: IFC Forward Swing

Forefoot/Rearfoot:

In the first minute, the percentage of IFC points in the forefoot region varied significantly across treatments, with NI showing the highest rate (10.3%), followed by FES (13.2%) and cfAFO (3.3%) ($p < 0.05$). These initial values highlight that the NI condition presented the highest likelihood of foot clearance occurring in the forefoot region, suggesting potential gait instability and a higher risk of tripping and falling.

All treatments showed an increase over time in the likelihood of the lowest point occurring in the Forefoot region during the last third of forward swing phase. During this part of swing, just before the highest point in the toe trajectory, the limb is approaching initial contact and a toe-down position is undesirable. The interventions appear to mitigate the extent of this shift. Both FES and cfAFO showed significantly lower probabilities of forefoot IFC occurrences compared to NI ($p < 0.05$), particularly in the later minutes of the test. This suggests that the interventions are effective in maintaining more stable and safer foot clearance patterns over time. cfAFO demonstrated the lowest tendency for a toe-down position throughout the 6MWT, with its greatest probability occurring at minute 5 at a very small 5.9%. This, compared with 21.8% for FES at minute 5 and 24.8% for NI at minute 6, support the cfAFO as providing the greatest control of foot-drop during this last portion of forward swing. Because of the rigid nature of the cfAFO, this result is not surprising; any occurrence of toe-down positioning at this late portion of swing would not occur from a change in ankle angle , but due to kinematics of the more proximal joints.

Medial/Lateral:

The results of this study reveal significant differences in the percentage of medial IFC point locations between all three conditions across the 6MWT. In minute 1, the percentage of medial IFC location was notably lower in the NI condition compared to both interventions, with FES showing the highest percentage and NI the lowest ($p < 0.001$). This trend persisted throughout the entire duration of the trial. Clinically, it is common to observe an increase in swing-phase ankle inversion in populations with CNS-caused foot-drop, often due to the presence of muscle spasticity. This may contribute to our finding that the NI condition shows a on average a lower percentage of IFC points in the medial region, as their ankle may be in a relatively inverted position. We must also consider that our results do not specifically reveal ankle position, but foot relative to ground, so it is possible that this result is due to altered hip kinematics. The observation that the highest percentage of medial mFC points occurred in the FES condition may be due to the difficulty in achieving desired frontal plane kinematics with the use of FES. A slight rotation of the FES cuff will change the degree of invertor/evertor activation at the ankle, sometimes causing a deviation from the desired neutral frontal plane positioning. In this study, the orthotist and physical therapist worked with participants to minimize this effect, though the dependence of the FES on user's neuromuscular response allows for variability

Probability of the mFC Location Occurring in the Forefoot Region

The probability of the mFC point occurring in the forefoot region was significantly lower for both FES and cfAFO interventions compared to the NI condition ($p < 0.0001$). The probability values across the minutes showed consistent trends, with NI consistently having the highest probability of forefoot mFC points, followed by cfAFO and FES. Notably, the comparison between cfAFO and FES did not yield a significant difference (p=0.9), indicating that both interventions are comparably effective in altering the mFC location from the forefoot.

However, we observed different trends when comparing changes over time in the probability of IFC and mFC locations occurring at the forefoot. The IFC measure showed an increasing trend in the percentage of points occurring in the forefoot region over the duration of the 6MWT across all conditions, with the NI condition exhibiting the most pronounced increase (from 87.6% to 90.9%). This suggests that as participants become fatigued, their average foot clearance during the forward swing phase shifts towards the forefoot, likely reflecting compensatory gait changes due to fatigue. In contrast, the AFC measure did not show consistent trends of change across minutes in the probability of forefoot occurrences.

These findings indicate that while the overall foot clearance (IFC) during the swing phase became increasingly forefoot-dominant with fatigue in both intervention conditions, the absolute lowest point of foot clearance (AFC) did not consistently follow this pattern. Fortunately, both interventions had a significant impact in lowering the probability of forefoot location throughout the 6MWT, in comparison to NI. Altogether, these findings suggest that the interventions might be particularly effective in preventing the most critical, lowest foot clearance points from occurring in the forefoot, even if the average clearance trends towards the forefoot over time.

Effect of Intervention on Timing of mFC Point within Forward Swing

Both interventions had a significant impact of the timing of mFC, moving the occurrence of the absolute lowest point within each forward swing later by around 5%. The cfAFO showed the greatest consistency of timing of mFC, through a much smaller coefficient of variation than FES or NI. This reduction of variability is probably attributable to the rigid structure of the cfAFO vs. the FES, whereas FES and NI both rely on neuromuscular performance, which can be highly variable in persons with CNS disorders. Theoretically, muscle fatigue may also play a greater role in foot/ankle kinematics in both FES and NI conditions due to this reliance on the user's neuromuscular response.

Effect of Interventions on Fatigability

Group averages for each intervention show an overall decline in walking distance from first to last minute of the 6MWT. However, while several participants met the cutoff score of -15% set by Leone et al for walking-related motor fatigue, no group averages met this cutoff. A DWI between -5% and -15% is considered to indicate some level of fatigability. By Leone's standards, the NI and FES groups did, on average, demonstrate some decline in walking performance, indicating an effect of fatigability.

Participant Preferences

There were no consistent trends across participants in their preferences compared to best interventions over selected outcome measures in Table 8. Four of eleven subjects preferred FES. Interestingly, only four of eleven participants preferred the device that held the best score on the majority of the six outcome metrics included in this table. All of these participants selected the cfAFO.

"Best" Interventions

When considering the six objective outcome metrics of (Table 8), the cfAFO emerged as the objectively "better" intervention in several categories. The cfAFO was associated with the greatest 6MWT distance and lowest % of IFC in the forefoot in 8 of 11 cases. It was associated with the highest AFC, highest averaged mFC, most favorable AFCΔ, and most favorable DWI in 6 of 11 cases. The interpretation that cfAFO is "better" due to these metrics is, of course, subject to the assumption that any increase in foot clearance is desirable. This assumption corresponds to literature relating low clearance to falls and foot scuffs, but there may be a limit, and it is possible that there are reasons to argue the contrary that have not been considered here. Furthermore, the right clinical selection is not solely determined by objective performance metrics; appearance, comfort, weatherproofing, cost, compatibility with footwear and clothing, and other concerns are also important in determining the right device for an individual.

The preference for either cfAFO or FES over NI by most participants underscores the subjective value of both devices in improving walking ability. These preferences, along with objective measures of gait improvement (6MWT average mFC, mFC location), support the use of such devices in the management of foot drop for PwMS. Individuality of participant responses holds implications for current clinical practice and for future study.

Clinical Implications

Given the lack of clear correlations between preference and various outcome measures included in this study, and the high variability in participant performance on various outcomes, an individualized approach to selecting the most appropriate foot-drop device is supported. Patient preference should be a major consideration in clinical decision-making. **However, there are several points in favor of further development of this relatively low-cost methodology for assessing foot clearance.** First, studies have shown that patient satisfaction is enhanced when provided with objective information to aid in decision making by their health care provider. Second, when considering that many individuals with foot drop have concurrent deficits in sensory systems including proprioception, our methodology may provided specific objective information that is not naturally perceived. For instance, not every patient will perceive changes in their frontal plane foot posture that we have shown can be more variable with an FES device than with a cfAFO, and such changes may be difficult for the orthotist or physical therapist to notice. Our method could help provide the user and their clinician with an understanding that the FES setup may require further adjustment, or that it is not the best solution to reduce risk of falls for that individual.

Further Research

Further research should examine larger populations and expand to populations with other root causes of foot drop, such as stroke, traumatic brain injury (TBI), or peripheral nerve damage. These patient populations are highly variable in ability and impairment profiles, which means that individual differences can significantly affect study outcomes. Applying this method to larger and more diverse populations will help generalize our findings to the broader populations and improve external validity of findings. Such exploration will provide the opportunity to refine our methods of assessing foot posture and improving its clinical utility.

Limitations

This study has several limitations that should be considered when interpreting the results. First, the sample size is relatively small, limiting the generalizability of the findings to the broader population of individuals with Multiple Sclerosis and foot drop. Second, this study lacked a control group to provide comparisons for novel metrics such as those based on foot clearance, including mFC region analysis, timing of mFC within forward swing, and coefficient of variation of mFC timing. Despite this, the crossover design enabled a comparison of participants to their own baseline metrics which does improve rigor by allowing a within-subject analysis and some extent of control for individual variability.

Another possible limitation is the two key interpretive assumptions regarding foot clearance: (1) that higher clearance is better, and (2) that increasing magnitude of improvement from NI is always meaningful. For assumption 1, an increase in clearance to values higher than what is observed for typical gait (1-2 cm for mTC) could potentially cause reduced economy of walking or unwanted stress to the more proximal joints and soft tissue structures in doing so. For assumption 2, some individuals may have relatively mild foot drop and therefore may not have much room for improvement. For such individuals, a large magnitude increase in foot clearance metrics from NI to one of the interventions may reflect a change that is larger than what is needed, leading to problems as mentioned under assumption 1.

Conclusions

In this study, we investigated the effects of carbon-fiber ankle foot orthoses (cfAFO) and functional electrical stimulators (FES) in foot clearance and fatigability during a 6-minute walk test in persons with Multiple Sclerosis and foot drop. Both interventions significantly increased mFC and AFC compared to no intervention, demonstrating their effectiveness in improving foot clearance and potentially reducing the risk of tripping and falling. The cfAFO intervention provided a greater increase in mFC compared to FES, suggesting that it may be more effective in enhancing foot clearance. However, this study also revealed significant variability in individual responses to the interventions, indicating that patient preferences along with objective outcomes should be considered.

Chapter 7: Summary

The purpose of this thesis was to explore mechanical underpinnings of postural control in gait and in standing balance in both health and neurologic injury, with the aims of advancing understanding of motor control and developing clinical application of novel metrics to assess and treat related deficits. We examined the relationship between task mechanics and motor control strategy through the Intersection Point (IP) and Xi, which describe foot-ground force interactions during standing and walking, respectively, and Minimum Foot Clearance (mFC), which provides a comprehensive view of whole-foot kinematics during swing phase.

Our work has further established the utility of θF-related metrics (IP and Xi) for assessing difference in motor control strategy of upright posture in standing and stance-phase of gait. Despite CNS damage, the presence of IP strategy persists in posts-stroke hemiplegia, but there exists significant differences in strategy between the paretic and non-paretic legs. We have demonstrated the IP to be a reliable and valid metric for measurement of postural control, a critical step in its path toward clinical translation.

Additionally, we have established that the Xi is sensitive to gait perturbations, likely reflecting subtle changes in motor control strategy necessary to preserve upright balance. Further, the Xi appears to adapt similarly for each leg when faced with a perturbation that impacts the body uniformly, such as posterior resistance force, and independently for each limb when presented with a perturbation that impacts each leg differently, such as walking on a split-belt treadmill.

We developed and piloted a novel method using IMUs and 3D scans to assess mFC in individuals with foot drop due to Multiple Sclerosis (MS). The study compared the effectiveness of two interventions, carbon-fiber ankle-foot orthoses (cfAFO) and functional electrical stimulators (FES) in improving foot clearance and reducing fall risk. Both interventions significantly increased mFC height compared to no intervention, with cfAFO showing a greater impact. Further, the cfAFO was shown to be more effective at reducing a toe-down ankle posture in late swing than the FES device, with both showing improvements over no intervention.

The metrics developed, validated, and applied in this thesis have significant potential for clinical translation. The IP and Xi metrics offer unique perspectives on motor control impairments and could enhance the precision of balance and gait assessments. The mFC method provides a detailed analysis of foot clearance, delivering crucial insights that can inform and improve clinical decisions among orthotic devices for preventing trips and falls in populations with gait impairments. Integrating these metrics into clinical practice could lead to more targeted and effective rehabilitation strategies, ultimately improving patiefnt outcomes.

Future Work

While this dissertation work has made several contributions to understanding and assessing postural control and gait mechanics, several areas warrant further exploration. Establishment of psychometric properties and greater inclusion of groups with diverse neurologically-based gait dysfunction would help generalize findings and refine these metrics for broader clinical applications.

First, while we established several psychometric properties of IP, there is a need for exploration of test-retest reliability over longer, clinically relevant time periods (weeks-months), and Xi should be subjected to similar studies. Test-retest reliability data over such periods has already been collected for IP, and this data will soon be analyzed.

Future work will include analyzing the regulation of Xi during single limb stan ce in adults with post-stroke hemiplegia. This analysis will be compared to results from healthy controls, providing further insights into the unique motor control strategies employed by individuals with hemiplegia.

Understanding the precise implications of changes in IP and Xi locations is essential before these metrics can be developed into treatment targets. For instance, it is currently unclear why the paretic limb experiences a lowering of IP height compared to controls, while the non-paretic limb shows a, possibly compensatory, increase. Gaining a deeper understanding of these changes will be critical for developing effective interventions aimed at restoring balanced postural control.

The novel method for assessing Minimum Foot Clearance (mFC) also warrants further study. Applying this method to larger and more diverse populations with neurologically based gait dysfunction will help generalize our findings and refine the analysis approach. Presentations in such populations are inherently heterogenous, and comprehensive analysis may reveal critical patterns or anomalies that are not evident in smaller groups. Additionally, exploring mFC dynamics in different contexts, such as varying walking speeds or terrains, can provide deeper insights into adaptive motor control strategies and their limitations in neurologically impaired individuals. This research has the potential to inform the development of more effective rehabilitation protocols and assistive technologies tailored to individual needs.

Real-world monitoring of foot clearance using the developed IMU and 3D scanning method could provide continuous data on gait and postural control in daily life. Such longitudinal data would be invaluable in understanding how interventions impact gait mechanics over time and in naturalistic settings. The BADGER Lab Real Life Wearables team is currently working on such an analysis from our Multiple Sclerosis cohort.

Overall, this research highlights the importance of integrating mechanical engineering principles with clinical insights to address complex issues in neurorehabilitation. Future studies should continue to build on these methodologies, aiming to translate them into practical tools for clinicians. By doing so, we can move closer to developing targeted, evidence-based interventions that significantly improve the quality of life for individuals with neurologic-based motor impairments.

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Appendix A

Double inverted pendulum simulation to relate joint torques to zIP and body kinematics. Mechanical simulation of the human as a two-segment inverted pendulum relates relative joint torques, body motion, and the foot force patterns described by zIP. Using anthropometric parameters (Shiozawa et al., 2021) a sagittal-plane model consisting of a legs rigid segment and a head-arms-trunk rigid segment was driven by a time ramp (0–10 Nm) of proportional hip and ankle torques to produce IP F behavior. Numerical simulations of 0.1 s at 200 Hz started with the segments vertical and at rest. The variation in xCP and Fx/Fz produced an IP (Gruben and Boehm, 2012). The full range of hip to ankle torque ratios were evaluated (n = 720) including pure ankle torque, pure hip torque, agonistic, and antagonistic ratios. The zIP is reported for selected torque ratios of interest (Fig. 3.3). Note that this model does not consider stability but only assesses the mechanical response to brief torques applied to a stationary vertical posture.

The simulation results indicate that a range of Thip/Tankle from -0.47 to 0.5 is required to produce the observed zIP range of 0.3 to 1.1 (Fig. 3.3). Within that range note that the behaviors of supra-ankle rigid body motion (ankle strategy) and of no ankle motion (exclusive hip strategy) each require significant torques at both joints (Thip/Tankle = 0.25 and 0.33, respectively). The observation of higher zIP values at low frequencies is likely due to the higher inertial properties (Creath et al., 2005) of the motions produced by the agonistic T ratios, compared to the antagonistic ratios, and thus the need to prolong the activation of high zIP coordination to achieve sufficient body displacement in the presence of high inertia. Conversely, the body motions where the segments rotate in opposite senses have lower inertial properties. Thus, they respond more quickly to torque inputs and require only brief (high frequency) inputs. Note that the Thip/Tankle = ∞ (pure hip T) has a zIP at the height of the ankle which is above the floor (floor z = 0).

Appendix B: zIP Description and Methods

In the sagittal plane, F orientation is approximated with the ratio Fx/Fz where Fx is the anterior-posterior component and Fz the vertical component. The anterior-posterior location of Fz is xCP (relative to some arbitrary fixed reference location). Within narrow frequency bands (~0.2 Hz width) of the Fx/Fz and xCP waveforms, a nearly linear relationship emerges between Fx/Fz and xCP (Figure A1). That relationship is geometrically equivalent to the F lines-of-action across time passing near a stationary point in space referred to as an intersection point (IP) [21] (Figure A2). The IP height (zIP) varies systematically with the band frequency such that zIP is above head height at low frequencies and around knee height at higher frequencies [21] (See example zIP vs frequency curve in Figure A3).

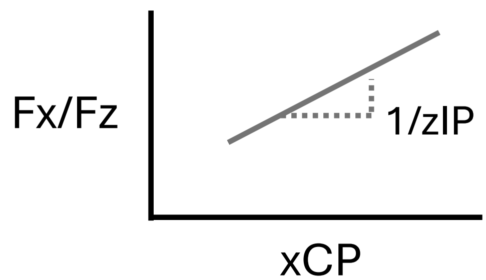


Figure B1: Linearity between Fx/Fz and xCP is equivalent to F having an IP with height zIP

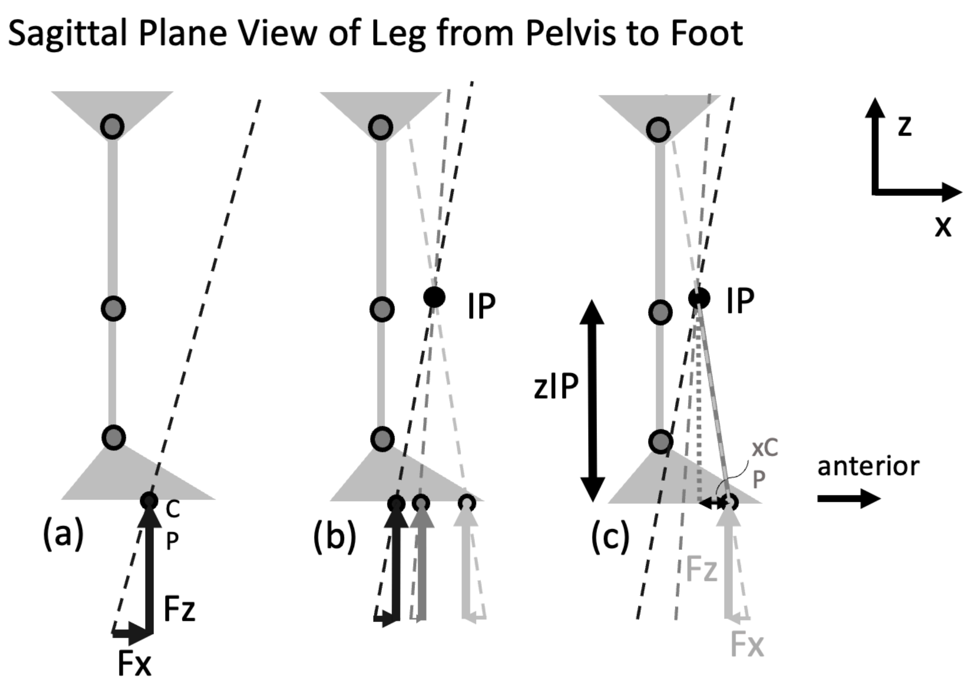
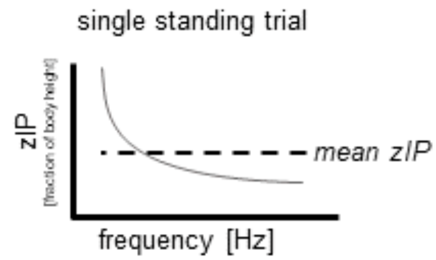


Figure B2: Across time, foot forces can have a geometrical property such that their lines-of-action pass near a point, called the intersection point (IP). (a) Resultant F line-of-action shown at the CP at a time instant. (b) Changes in CP location and F orientation (Fx/Fz) over time with an IP. (c) The ratio Fx/Fz is equal to the ratio xCP/zIP at each time instant.



Participant	**Condition**	**Repetition**		
1	1	1		
1	1	2	*intra-condition mean zIP*	
1	1	3		
1	2	1		
1	2	2	*intra-condition mean zIP*	
1	2	3		*summative zIP*
1	3	1		
1	3	2	*intra-condition mean zIP*	
1	3	3		
1	4	1		
1	4	2	*intra-condition mean zIP*	
1	4	3		
2	1	1		
...	...	...		

Figure B3: Organization of zIP variables. Each 60-second standing trial produced one zIP versus frequency curve. From that curve a *mean zIP* value characterized that trial. *Mean zIP* from the three repeated trials were averaged to produce the *intra-condition mean zIP*. Those were summed to provide *summative zIP* that represented the participant across all 12 trials.

Appendix C: Detailed Description of Measures

Short Falls Efficacy Scale-International (SFES-I): (Kempen et al 2008)
The SFES-I is a reliable and valid 7-item questionnaire designed to measure a person's concern about falling while getting dressed or undressed, taking a bath/shower, getting on/off a chair, walking up/down stairs, reaching for something high/low, walking down a slope, and going to a social event. For each question, respondents indicate whether they are "not at all concerned" = 1 point, "somewhat concerned" = 2 points, "fairly concerned" = 3 points, or "very concerned" = 4 points. Thus, the maximum score is 28 and higher scores indicate greater concern for falling.

International Physical Activity Questionnaire-Short Form (IPAQ-SF): (Craig et al., 2003)
The IPAQ-SF is a reliable and valid measure of physical activity of four intensity levels: vigorous (e.g., aerobics), moderate (e.g., leisure cycling), walking, and sitting. Respondents report the frequency and duration of engaging in these different intensity levels of activity in the prior seven days. Scores are reported as low, moderate, and high activity level.

Trail Making Test (TMT): (Bowie & Harvey, 2006)
The TMT is a measure of visual scanning and working memory that has been shown to have good inter-rater reliability in adults, though this test is susceptible to practice effects. It has also been shown to be valid in adults [30]. The TMT is divided into part A (TMT-A), which assesses rote memory, and part B (TMT-B), which is a test of executive function. In part A, participants are asked to draw a line to consecutively connect a series of numbers (1 to 25). In part B, participants are required to draw a line to connect numbers and letters in an alternating progressive sequence (1 to A to 2 to B and so forth). Each test is timed and increased ratios of TMT-B relative to TMT-A are thought to indicates cognitive impairment; however, cutoff scores have not been established due to lack of specificity, and times > 273 seconds on the TMT-B indicate cognitive impairment.

Mini-Balance Evaluation Systems Test (mini-BEST): (Franchignoni et al., 2010)
The mini-BEST is a multi-dimensional assessment of balance during standing and walking that has been shown to be reliable and valid in adults. The mini-BEST includes 14 tasks that each belong to one of four domains. Anticipatory balance (0 to 6 points) is tested during sit-to-stand, rise-to-toes, and standing on one leg. Reactive balance (0 to 6 points) is assessed during compensatory stepping forward, backward, and sideways to regain balance. Sensory contributions to balance (0 to 6 points) are evaluated during quiet standing on the floor with eyes open and eyes closed, as well as while standing on an incline with eyes closed. Dynamic balance (0 to 10 points) is evaluated during walking with speed changes, walking with horizontal head turns, a pivot turn, stepping over obstacles, and while completing a Timed Up and Go with a dual task. Total scores range from 0 to 28 points with higher scores indicating better balance function.

Posturography Configuration: (Santos & Duarte, 2016).
The foam surface was a six cm thick foam block (Balance Pad; Airex AG, Sins, Switzerland). Participants stood quietly for 60 seconds barefoot on a force plate with their feet 10 cm apart and at an angle of 20 degrees from the midsagittal plane with their arms at their sides for each trial. During the eyes open conditions, participants fixated on a 5 cm round black target that was placed 3 m ahead at eye level.

The extent to which foam compresses could vary systematically with individual characteristics such as body mass and foot area. To assess this dependency, we performed stepwise linear regression for the impact of these factors on the greater trochanter marker height for participants in a parallel study (Koo & Li, 2016) that included motion capture enabling tracking of vertical location of anatomical landmarks. Four potential predictors of greater trochanter height relative to ground were analyzed: condition (foam vs. no-foam), body weight, overall height, and foot length. Body mass and foot length did not contribute significantly to the regression. There was a main effect of body height, as expected, coefficient = 0.57. The coefficient for foam/no-foam was 0.047 m which was used to shift zIP from force plate coordinates to foot sole coordinates.

Appendix D: Curriculum Vitae

JENNIFER N. BARTLOFF

9349 N. Lowell Avenue Skokie, IL | 574-302-2852 | bartloff@wisc.edu

EDUCATION

2024 PhD in Mechanical Engineering, University of Wisconsin-Madison
2021 Master of Science in Mechanical Engineering, University of Wisconsin-Madison
2009 Doctor of Physical Therapy, Northwestern University, Chicago, IL
2003 Bachelor of Science in Cognitive Science, Indiana University-Bloomington
Concentration in Neuroscience, Minor in Mathematics

RESEARCH EXPERIENCE

2019-2024 Research Assistant, Neuromuscular Coordination Lab, UW-Madison

Evaluation of Xi in Non-Neurologically Impaired and Post-Stroke Ambulators

- Conceptualization of secondary analysis of split-belt instrumented walking trials where a posteriorly directed resistance force was applied to participants' centers of mass
- Data analysis in R
- Currently drafting manuscript for results in healthy controls, analyzing post-stroke data

Test-Retest Reliability of the Force Intersection Point

- Conceptualization, protocol design, IRB approval, supervision, data collection, software design (Labview), data analysis
- Manuscript in preparation.

Translating and Rotating to Explain Motor Control of Balance with the Lower Extremity (TREMBLE)

- Design and construction of TREMBLE device
- Motor-driven force-sensing support surface that enables pivoting around the ankle joint of the standing human and translation along an anterior-posterior axis
- Device measures body posture via mechanical motion capture to facilitate closed-loop control of movement based on user behavior

2019-2024 Research Assistant, BADGER Laboratory, UW-Madison

Evaluating Mobility Interventions in the Real World

- Protocol Design for ClinicalTrials.gov registered project
- Project management including IRB approvals, participant recruitment, screening, scheduling, and testing
- Development of study-specific qualitative questionnaires

- Design of REDCap environment for project management
- As-needed phone support for participants utilizing Bioness functional electronic stimulator device
- Development of novel method for analyzing foot clearance using 3D scan and a single inertial measurement unit (IMU) mounted on the shoe
- Currently analyzing data and drafting two manuscripts for results, methods paper submitted in January to Scientific Reports

LICENSURES & CERTIFICATIONS

2019-Present Physical Therapist, State of Wisconsin License Number 14496-24
2018-Present Board Certified Clinical Specialist in Neurologic Physical Therapy
2015 **American Physical Therapy Association**
Advanced Credentialed Clinical Instructor
2015 Certified Bicycle Fitter, BikePT Gold Level, bikept.com
2012 LSVT BIG Certified Physical Therapist

CLINCIAL EXPERIENCE

2019-2022 Physical Therapist, UW Health, Madison, WI
- Inpatient rehabilitation, outpatient neuro, and acute settings

2010-2018 Physical Therapist**,** Asante Rogue Regional Medical Center, Medford, OR
- Specialist in neurologic rehabilitation across settings of outpatient, inpatient rehabilitation, and acute care
- Development of neurologic treatment program including guidelines for post-treatment follow-up and allocation of resources to maximize service to our community
- Patient education material development including stroke awareness, exercise principles for Multiple Sclerosis, and Parkinson's movement and exercise strategies
- Outreach and collaboration with local gyms and personal trainers for post-discharge continuity
-

MENTORSHIP

Summer 2021 INTEREGR 660: Research Mentor Training Practicum, UW-Madison
2021-Present Mentor for undergraduate students in BADGER Lab and Neuromuscular Coordination Laboratories, UW-Madison
2021-2023 Mentor, Department of Kinesiology KinEqt Program, UW-Madison
2020-2022 Mentor, Women in Scientific Education and Research (WISER), UW
2012-2018 Mentor to 1st year therapists, Asante Health System, Medford, OR

LEADERSHIP

2020-2021 Panelist, Women in Scientific Education and Research (WISER)
2017-2020 Team Captain, American Diabetes Association Tour de Cure, Team Invinsulin
2014-2015 Member, Lane Community College PTA Program Advisory Council, Medford, OR
2005-2006 Trip Leader, Outdoor Adventures, University of Michigan-Ann Arbor, MI

TEACHING EXPERIENCE & TRAINING

2023-2024 Teaching Assistant, Departments of Mechanical and Biomedical Engineering, UW-Madison, Madison, WI
- BME 315: Biomechanics, BME/ME 415: Biomechanics of Human Movement

Fall 2023 Guest Instructor, Doctor of Physical Therapy Program, UW-Madison
- PT 635: Motor Control Dysfunction I: *Technology in Neurorehabilitation of the Lower Extremity*

2019-2023 Teaching Assistant, Department of Kinesiology, UW-Madison, Madison, WI
- KINES 338: Anatomy Lab, KINES 318: Biomechanics of Human Movement
- Developed Lab Content including integration of OpenCap + OpenSim, EMG, force plates.

Fall 2021 Primary Instructor, Department of Kinesiology, UW-Madison, Madison, WI
- KINES 318: Biomechanics of Human Movement
- Developed course materials integrating evidence-based principles of scientific teaching and diversity, equity, and inclusion.
- Mentored teaching assistants

Spring 2021 INTEREGR 650: Graduate coursework in effective teaching strategies in the sciences, UW-Madison, Madison, WI

2012-2018 Clinical Instructor, Asante Health System, Medford, OR
- Mentored 10 Students of Physical Therapy from programs across the US

2015-2018 Faculty, Lane Community College, Eugene, OR
- Physical Therapy Assistant Program
- Kinesiology and Neurology lab instruction and curriculum development

PUBLICATIONS

Fehr KH, **Bartloff JN**, Wang Y, Hetzel S, & Adamczyk PG. (2024). Estimation of minimum foot clearance using a single foot-mounted inertial sensor and personalized foot geometry scan. *Scientific Reports*, *14*(1), 13640. https://doi.org/10.1038/s41598-024-63124-6

Bartloff JN, Gruben KG, & Grove CR. (2024). Reliability and validity of the force intersection point in the assessment of human quiet standing balance. *Human Movement Science*, *96*, 103239. https://doi.org/10.1016/j.humov.2024.103239

Bartloff JN, Ochs W, Nichols KM, & Gruben KG. (2024). Frequency-dependent behavior of paretic and non-paretic leg force during standing post stroke. *Journal of Biomechanics*, *164*, 111953. https://doi.org/10.1016/j.jbiomech.2024.111953

Fehr HK, **Bartloff JN**, Wang Y, Konieczka K, Mastej J, Adamczyk PG. Determining whole-foot minimum clearance by augmenting IMU trajectory with personalized 3D scans. *(In Review) Submitted to Scientific Reports, January 2024*

Bartloff JN. Brown DA, Moradian N, Ko M, Gruben KG. Independent Foot Force Regulation for Postural Stability During Walking. *(In Preparation, Target journal: PT Journal)*

Bartloff JN, Gruben KG, Grove CR. Reliability and validity of the force intersection point in the assessment of human quiet standing balance. (*In Review) Human Movement Science #HMS-D-23-00602*

Dutt-Mazumder A & Gruben KG. (2021). Modulation of sagittal-plane center of pressure and force vector direction in human standing on sloped surfaces. *Journal of Biomechanics*, *119*, 110288. **Acknowledged in publication** for enhancing the graphical representation of the study's outcome measure. https://doi.org/10.1016/j.jbiomech.2021.110288

Bishara et al. Similar Processes Despite Divergent Behavior in Two Commonly Used Measures of Risky Decision Making. J Behav Decis Mak 2009;22:435-454. Acknowledged in publication for data collection.

PRESENTATIONS

Bartloff JN, Fehr KH, Wang Y, Konieczka K, Mastej J, Cohen E, Adamczyk PG. A Mixed Methods approach to evaluating foot-drop devices in multiple sclerosis. Poster, American Physical Therapy Association Combined Sections Meeting. Boston, MA 2024.

Bartloff JN, Fehr KH, Wang Y, Konieczka K, Mastej J, Cohen E, Adamczyk PG. Pairing IMU trajectory with 3D foot scans for comparison of whole-foot minimum clearance among foot drop interventions in Multiple Sclerosis. Oral Presentation, Rocky Mountain Muscle Symposium Pre-Conference Summit. Canmore, Alberta, 2023.

Bartloff JN, Brown DA, Moradian N, Ko M, Gruben KG. Independent foot force regulation for postural stability during walking: Insights from force and speed perturbations. **Poster, Progress Clinical in Motor Control II. Chicago, IL, 2023.**

Bartloff JN, Fehr KH, Wang Y, Konieczka K, Mastej J, Cohen E, Adamczyk PG. Mixed-methods Approach to Foot-Drop Device Selection in Multiple Sclerosis: An Exploratory Study**. Poster, Consortium of Multiple Sclerosis Centers (CMSC). Aurora, CO, 2023.**

Bartloff JN, Nichols K, Ochs W, Fox J, Gruben KG. Parameterization of frequency dependent lower-limb coordination during human standing after stroke. Poster, Society for Neuroscience. San Diego, CA, 2022.

Gruben KG, Grove CR, **Bartloff JN**. Vision and proprioceptive effects on force intersection point in quiet human standing. Poster, Society for Neuroscience. San Diego, CA, 2022.

Bartloff JN. Force direction control after brain injury. Oral Presentation & Poster, Institute for Clinical and Translational Research Midwest TL1 Research Summit. Madison, WI, 2022.

Fehr KH, **Bartloff JN**, Wang Y, Mastej J, Konieczka K, Acasio JC, Knight A, Hendershot BD, Adamczyk PG. Toward evaluating prosthetic feet using real-world data: preliminary results on whole foot clearance and knee angle. Poster, Military Health System Research Symposium (MHSRS). Kissimmee, FL, 2022.

Fehr KH, **Bartloff JN**, Wang Y, Mastej J, Konieczka K, Adamczyk PG. Determining whole-foot ground clearance kinematics by augmenting IMU trajectory with personalized 3D scans. Poster, North American Congress of Biomechanics (NACOB). Los Angeles, CA, 2022.

Bartloff JN, Fox J, Gruben KG. Foot force vector intersection reveals shared control structure across tasks. Oral Presentation, Dynamic Walking Annual Conference. Madison, WI, 2022.

Fehr KH, **Bartloff JN**, Wang Y, Mastej J, Konieczka K, Adamczyk PG. Real-World Whole Foot Ground Clearance. Poster, Dynamic Walking Annual Conference. Madison, WI, 2022.

Grove CR, **Bartloff JN**, Gruben KG. Validation of the intersection point balance metric in adults. Poster Accepted**,** Academy of Neurologic Physical Therapy Annual Conference Virtual Conference, 2021.

Bartloff JN, Gruben KG. Improved metric of balance coordination during quiet standing. Poster, American Physical Therapy Association Combined Sections Meeting. Virtual Conference, 2021.

Bartloff JN, Grove CR, Gruben KG. Foot force control in gait and balance. Poster and Panelist, Institute for Clinical and Translational Research Symposium: Detecting and Managing Balance and Mobility Phenotypes Using Next Generation Technologies: Artificial Intelligence, Biomechanical Sensing, and Beyond. Virtual Conference, 2020.

Gruben KG, **Bartloff JN**. A novel measure of standing balance coordination demonstrates greater consistency than traditional measures. Poster Accepted, American Society for Biomechanics Annual Meeting. Virtual Conference, 2020.

Nichols K, Lefranc A, **Bartloff JN**, Gruben KG. Intermittent linearity of foot force direction vs center of pressure in quiet standing. Poster, Society for Neuroscience Annual Meeting Chicago, IL, 2019.

Bartloff JN, Bitting J, Lueke A, Sbertoli C, Sofen L, Walsh J, Johnston C, Hilliard MJ, Brown DA. After-effects of slow isokinetic walking speed on self-selected gait velocity in persons with chronic post-stroke hemiparesis. Poster, American Physical Therapy Association Combined Sections Meeting San Diego, CA, 2010.

SKILLS

Programming: R, Matlab, C, C++, Python, HTML, ROS, p5.js, GitHub, parallel programming techniques
Software and Tools: Opensim, Fusion 360, OnShape, Motive, REDCap, Adobe: Illustrator, Photoshop, Premiere Pro; Inkscape, OpenCap, Zotero, ResearchRabbit, PyChrono, Visual 3D, FEBio, Oculus Rift
Instrumentation: Force Plates, Optical & Mechanical Motion Capture, APDM Opal Inertial Measurement Units, K5 Metabolic Mask, MTS machines, XSense, Movella Dots, EMG, Raspberry Pi
Clinical Instrumentation: Handheld dynamometry, EMG biofeedback, Neurocom Balance Master, Bioness Functional Electronic Stimulation, Neuromuscular Electronic Stimulation
Fabrication: CNC router, wood/metal lathe, laser cutter, additive manufacturing, general metal/wood shop equipment, soldering and circuit design

OUTREACH

2019 — Exhibitor, Engineering Expo, UW-Madison

- Assisted in preparation of and presentation of exhibit representing the Neuromuscular Coordination Lab
- Outreach to local junior and senior high school students

September 2022 — Speaker, Adapted Fitness Program, UW-Madison
"Opportunities for Research and Careers in Elite & Recreational Para-sport"

September 2019 — Speaker, Pre-Physical Therapy Club, UW-Madison
"My Career in Physical Therapy and Research"

Fall 2018 — Volunteer Instructor, Talent Maker City, Talent, OR

- STEAM workshop for underrepresented groups
- Kinesiology and Neurology lab instruction and curriculum development

Fall 2017 Media Appearance, "Better Health" Segment for KOBI, Medford, OR

- Topic: Update on the Asante neurologic PT program

2016 Speaker, Parkinson's Resources of Oregon, Medford, OR
"The role of amplitude-based exercise in treatment of Parkinson's Disease"

SERVICE

2022-Present Item Writer, American Board of Physical Therapy Specialties
2020-Present Classifier, Union Cycliste Internationale (UCI) Paralympic Cycling
Highlights: 2024 Para Track World Championships, Brazil
2023 ParaPanAmerican Games, Chile
2022 UCI World Cup Quebec City, Quebec
2018-Present Classifier, USA Cycling (USAC) Paracycling
2013-2019 Volunteer, Oregon Mountain Bike Association Trail Building and Repair
2017 Assistant Coach and Official, Special Olympics of Southern Oregon Bocce, Medford, OR
2009 Volunteer, Sky Rise Chicago Adapted Wheelchair Race, Chicago, IL
2009 Volunteer, Illinois State Cyclocross Championship, Chicago, IL
2007 Medical Volunteer, Chicago Marathon, Chicago, IL
2003-2004 Volunteer Mechanic, Kansas City Bicycle Project, Kansas City, MO

HONORS AND AWARDS

2023 Promising Investigator Award, Rocky Mountain Muscle Symposium Pre-Conference Summit
2023 Travel Award, Rocky Mountain Muscle Symposium Pre-Conference Summit
2021 Elizabeth M. Roberts Scholarship, UW-Madison
2021 Conway Fellowship, UW-Madison

- *Turned down due to NIH TL1 Funding*

2021 Carns, Cronin, Glassow Scholarship
2019-2022 National Institutes of Health TL1 Pre-Doctoral Award, Team Sciences Trainee
2019 & 2020 Blanche Trilling Fellowship
2019 Grand Challenges Seed Grant Competition Winner
2019 New Graduate Student Scholarship, Dept. of Kinesiology
2015 Parkinson's Resources of Oregon Travel Grant
2015 Overall Series Champion, Portland Cyclocross Crusade
2015 Oregon Bicycle Racing Association Best All Around Rider, Cyclocross Masters Women
1999-2003 STARS Undergraduate Research Scholar, Indiana University-Bloomington
2002 Honors College Creative Activity Grant, IU-Bloomington
1999-2001 Resident Scholar, IU-Bloomington
1999 Tiscornia Scholar, St. Joseph, MI
1999 Jay Gargano Memorial Scholarship, Stevensville, MI