Understanding the role of lower limb kinetic adaptations in dynamic stability during novel forward walking

by

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Abstract

Introduction

During asymmetrical gait perturbations, adaptive alterations in spatiotemporal (i.e., step width & length) and kinematic parameters (i.e., margins of stability (MoS)) have become an important means to probe the mechanisms of stability control. Recent work has linked eccentric ground reaction force (GRFnet) control to ML instability during normal and fast paced walking, potentially yielding insight into proactive and reactive mechanisms of stability control. Using a split-belt treadmill, where gait perturbations can be administered by decoupling individual belts, this study sought to examine adaptations in the kinetic mechanisms underlying stability in gait, which have yet to be examined. The timing and magnitude of the angle of GRFnet eccentricity (θd) were examined during the double support phase to better understand how younger adult individuals modulate forces in relation to situational demands to maintain stability.

Objective

To examine timing and magnitude of potential proactive and reactive control indices of stability (i.e., initial (P1) and later phase (P2) GRFnet eccentricities) to better understand how individuals modulate forces to maintain dynamic stability in the presence of a novel gait pattern.

Methods

Whole-body kinematic and kinetic data were collected from twenty-eight young adult participants. Participants completed a 15 min split-belt protocol in which the left belt (0.75 m/s) was slower than the right belt (1.5 m/s). This continuous perturbation was used to provoke instability in which adaptation in control mechanisms could be observed during early adaptation (EA) and late adaptation (LA) time points. Step width and margins of stability were calculated, and specific focus was placed on the on angle of divergence of the net ground rection force. Two-way repeated measures ANOVAs were used to assess adaptation across time points and between individual limbs to further our understanding of dynamic stability.

Results

During EA participants exhibited conservative control strategies as observed by increased MoS coupled with decreased initial GRFnet (P1) eccentricity and increases in later GRFnet (P2) eccentricity, while no differences in timing were observed. Additionally, step-to-step variability increases in MoS, P1, and P2 magnitude were noted during EA. During LA individuals exhibited similar control strategies relative to baseline, demonstrated by reduced MoS and increases in P1. Further, decreases in step-to-step variability of stability control parameters were also noted during LA.

Discussion

Findings suggest that changes in spatiotemporal and force related control mechanisms during a continuous whole-body perturbation are requirements of stability preservation. Further, our results suggest that some ML control parameters exhibit adaptive changes, that is, over time there is a lesser reliance on reactive control measures – these results may be exclusive to a population which can offset instability by allocating control appropriately between limbs to achieve suitable maintenance of dynamic stability. Further work is necessary to examine the potential for such adaptive changes among older adults.

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Chapter 1: Introduction

In Canada the older adult population, that is, adults 65 years and older currently make up 16.9% of the total population (Statistics Canada, 2016). Data from the previous Canadian Census Program would show that the older adult population is climbing, with a 2.1% percent increase between the two nationwide questionnaires (Statistics Canada, 2011). With the 2016 Census Data stating that 66.5% of the nation's population falls within the 15 to 64 group; one may be able to infer that future data may continue to show increasing rates of older adults in Canada (Statistics Canada, 2016). Further, as the older adult population increases so will the economic burden. Of all the age groups, older adults are among the most at-risk category for injury due to age-related declines in overall health. A report from the Canadian Community Health Survey stated that injuries are the leading cause of hospitalizations among all population groups, but falls remain the top cause of older adult injury-related hospitalizations (Stinchcombe et al., 2014). Falls and fall related injuries not only harm the affected individuals, but also put strain on the loved ones, friends, and ultimately the health care system. It is estimated that the economic burden associated with older adults and fall-related injuries is ~$2 billion annually (Stinchcombe et al., 2014).

Consider that in the older adult population falls resulting in physical trauma and daily activity restrictions are the leading causes of morbidity (Tinetti et al., 1993). Falls are among the top public health concerns in relation to the elderly. An explanation for this may be due to the lasting effects of fall related trauma. If an older adult experiences a fall related injury that affects physical performance, this may affect the type of activity individuals do post-injury. A review of the relationship between fear of falling and associated activity restriction suggests that fear of falling leads to a risk to lowered mobility and activities of daily living (Denkinger et al., 2015). Thus, a cycle is formed; those who experience a fall or fall resulting in injury become fearful of falling again. This may continue to affect how individuals go about their daily routine, ultimately, reducing daily living activities, leading to further mobility challenges and an increased risk of falls.

Regarding the risk of falls in older adult populations, one must consider situations that put individuals at an increased risk of falls. Further, we must engage with potential outcomes on specific falls. Work characterising falls among older adults suggest that the majority of falls occur indoors, with only 26% of falls occurring outdoors (Stevens et al., 2014). Furthermore, falls can be categorized by the direction in which they occur; in older adult populations falls in the forward direction are among the most common, followed by falls to the side and backward direction which can be more greatly associated with hip fractures (Greenspan et al., 1998; Robinovitch et al., 2013; Schwartz et al., 1998; Stevens et al., 2014). These results may be a product of the activities in which older adults are most likely to experience a fall. As such, research suggests that most falls occur during common daily activities such as forward walking, sitting down, and standing as the individuals are more likely to experience loss of balance or a fall related to slipping and tripping (Robinovitch et al., 2013; Stevens et al., 2014). Additionally, there is evidence to suggest that women are more at risk of balance disturbances resulting in falls than are men (Greenspan et al., 1998; Robinovitch et al., 2013; Stevens et al., 2014).

Among all age groups walking may be the most important form of locomotion, as most individuals use walking to traverse through everyday life. As humans age our risk for falls and fall related injuries increase, but the importance of walking remains unchanged, thus, a spotlight should be cast on the underlying mechanisms behind the age-related increase in fall risk to ensure that the aging population can safely continue moving independently. Work addressing older adult walking strategies found that as we age, we are more likely to walk slower and with an increase in step frequency (Kang & Dingwell, 2008). These adaptations to the gait cycle have been suggested to be a product of a reduction in muscle strength (DeVita & Hortobagyi, 2000), and a decrease in flexibility (Kerrigan et al., 2001). Further, reductions in muscular strength are correlated to decreased functional walking performance in relation to balance control, therefore, these changes in locomotion may support the change in movement strategy in which older adults redistribute work from distal joints to proximal (DeVita & Hortobagyi, 2000; Kerrigan et al., 2001; Moxley Scarborough et al., 1999). Although older adults adopt a slower walking pattern as a function of declines in muscle strength, it has also been suggested that the decrease in walking speed can aid in the increasing of local instability, even when an increase in gait variability is present (Hak et al., 2012; Kang & Dingwell, 2008). The increase in local stability may decrease the likelihood that an internal perturbation is responsible for the increased fall risk but highlights the importance of an individual's ability to deal with external forces acting on the body.

Based on the mathematical modeling it has been suggested that bipedal stability is both actively and passively controlled. During forward walking, sagittal plane stability is governed by passive dynamics, that is, push off during the gait cycle will have little effect on step length, but will increase COM velocity. At the time of heel strike, the increase in velocity will allow for dispersion of the heel collision, offsetting nominal energy and to counter the disturbance to stabilize the gait cycle (Donelan et al., 2004; Kuo & Donelan, 2010). Whereas frontal plane stability during walking is suggested to be actively controlled by lateral foot placement. During above ground and treadmill walking it has been reported that step width variability remains higher than step length variability (Bauby & Kuo, 2000; Owings & Grabiner, 2004). This may support the notion that maintenance of lateral stability requires active control based on sensory feedback; whereas sagittal plane stability is passively regulated - the need for active/reactive control, or perhaps even sensory input, may be minimal. Additionally, the active recalibration of margin of stability parameters in response to perturbations may be governed by foot placement strategies or lateral ankle strategies by which, during push off, the regulation of ankle inversion/eversion is used to modulate the body's trajectory to preserve mediolateral stability in subsequent steps (Buurke et al., 2018; A L Hof et al., 2005; At L. Hof, 2008; Reimann et al., 2017). In relation to older adult movement patterns, it has been documented that older adults redistribute joint forces from the ankle to the hip as a main mechanism to control lateral COM movement, that is, larger amounts of gluteus medius muscle activation take place during mid- and terminal stance phases; with larger lateral hip moments lasting longer throughout the gait cycle in older adult populations, (DeVita & Hortobagyi, 2000; Kerrigan et al., 2001; Moxley Scarborough et al., 1999; Schloemer et al., 2017). Regulation of foot placement strategies have been more extensively studied in older adult populations because foot placement is determined by changes in hip adduction/abduction during the swing phase of walking. Moreover, it has be speculated that changes in inversion/eversion during foot roll-off may require more force generation (Reimann et al., 2017), which refutes the idea that humans complete tasks based on metabolic work efficiency and instead suggests that preservation of stability may take priority.

During assessments of standing stability, one remains 'stable' if their horizontal centre of mass projection lies within their base of support. However, in dynamic situations such as human walking this modelling does not apply as it does not consider the horizontal velocity of the COM. Hof et al. (2005) proposed the margin of stability to measure dynamic situations, that is, a variable which uses the velocity corrected COM (XCoM) and an individual's BOS in any point in time. Further, MoS is related to the impulse needed to cause instability (Hof et al., 2005). Consider during walking each step may cause a destabilizing impulse during each heel strike, therefore, foot placement may be a mechanism used to control MoS (Hof, 2008). Consequently, the changes in foot placement may be a mechanism aimed at preserving a minimum MoS needed to remain in a stable position. During overground and treadmill walking, increases in ML step variability and minimal changes in average MoS are observed, therefore, describing a scenario in which individuals use foot placement as a means to modulate MoS such that the system can remain stable (Bauby & Kuo, 2000; Owings & Grabiner, 2004; Rosenblatt & Grabiner, 2010). Further research addressing MoS between steps may offer more information regarding active control of stability.

During normal gait it has been proposed that variation in gait parameters both between and within limbs reflects the capacity of the balance control system to maintain some level of optimal functionality (Harbourne & Stergiou, 2009). Although variation among walking gait requires some level of variation to ensure adjustments to external and internal stressors as they arise, it has been shown that between-limb coordination may also play a role in the maintenance of stability during gait. The study of interlimb coordination research allows for further investigation of the neuromuscular control system and its role in actively modulating gait parameters. In younger adult populations changes in joint coordination are seen by changes in continuous relative phase data at the hip and knee when adapting to changes in walking speed on flat surfaces (Chiu et al., 2014; Chiu & Chou, 2011; Hafer & Boyer, 2018). This may suggest a coordination strategy that favors changes in proximal joint dynamics rather than distal changes when adapting to gait stimuli. Furthermore, when walking mechanics are challenged as a function of walking surface, uneven walking surfaces led to younger adults exhibiting greater in-phase coupling at the knee-hip and ankle-knee (Ippersiel et al., 2021). This may suggest that as neuromuscular system becomes more challenged, the extent of variability and coordination strategies change, that is, more use of distal joints to preserve stability.

A recent shift addressing the mechanisms of stability control have introduced the use of split belt treadmills and asymmetric gait pattens to shed light on how individuals actively control stability through coordination strategies. The use of asymmetric gait patterns has shown adaptation strategies to help promote more dynamic stability. Consider the relationship between whole-body angular momentum and asymmetrical step length; individuals who do not suffer from neuromuscular impairments show no changes in whole body angular momentum as a result of changes in asymmetrical step parameters. Further, work has suggested that intersegmental limb coordination is altered during asymmetric gait, and these changes are a result of a shift in lower limb momentum to reach a target symmetry yet no changes in whole body angular momentum are observed (Liu & Finley, 2020). This may suggest that the central nervous system governs whole-body angular momentum by allowing variability in step parameters and lower limb coordination (Liu et al., 2018; Liu & Finley, 2020).

Walking is among the most fundamental locomotor tasks needed to execute tasks of daily living; however, little is known about the forces and moments that account for adaptations in coordination during normal gait patterns to preserve stability. Previous work has instead used kinematic measures to establish variation among coordinative strategies, that is, differences in relative phase and joint angles as a means to measure coupling between and within limb movement in healthy gait patterns (Chiu et al., 2014; Chiu & Chou, 2011; Hafer & Boyer, 2018; Ippersiel et al., 2021). Similar outcome measures have been used to address coordination in novel asymmetrical gait patterns to address coordination adaptations (Haddad et al., 2006). Literature addressing coordination and asymmetrical gait patterns has often omitted kinetic outcome variables as a measure to quantify adaptations. As a result, we have knowledge of the behavioural outcomes, but lack understanding of the kinetic mechanisms responsible for such kinematic and temporospatial adaptations.

The current work aimed to further information surrounding dynamic stability during forward walking by using kinetic outcome measures – specifically, the magnitude, orientation, and timing of the net ground reaction force (GRF_{net}). Although magnitude of GRFnet has been linked to lower limb strength (King et al., 2012), the orientation of initial (P1) and later (P2) phase GRFnet eccentricities (i.e. angle of the GRFnet relative to the COM) have been suggested to be linked to their respective proactive and reactive control mechanisms needed to restabilize COM kinematics through the conservation of angular momentum (Singer et al., 2014, 2016). Further, the timing of applied forces may also help uncover whether changes in stability control are linked to an individual's ability to sense excessive COM motion. Force related measures of instability, coupled with temporospatial, and kinematic step characteristics (step width & length; ML MoS) further aid in the understanding of how external gait perturbations elicit changes in force output that are responsible for changes in step characteristics.

This work may provide information on a novel approach to addressing dynamic stability as a product of interlimb kinetic coordination. Additionally, this work has the potential to explain some of the kinematic adaptations seen during normal walking; that is, shifts segmental coupling. Going forward the information put forth in this thesis will provide a literature review which outlines the past and present work surrounding stability and coordination among adult populations. Moreover, the integration of different methodologies used to extract balance and stability measures pertinent to the proposed topic. Further, insight will be given regarding split-belt treadmills and the role of kinetic outcome variables when addressing stability related problems among adults. This thesis aims to act as a foundation for addressing kinetic mechanisms for stability control in the presence of continuous whole-body gait perturbations. This work may in fact provide information surrounding the importance of maintaining stability and will aid in the understanding of how the body prioritizes stability. Although investigating age-related changes in stability control is a pertinent line of inquiry, we must first understand situational adaptations in stability control that arise during continuous gait perturbations to aid in understanding age-related differences. As such, the future goal of our work will be to understand age-related differences in stability adaptations such that distribution of specific interventions can be used to augment dynamic stability in hopes of decreasing the risk of falls for those who are more at risk.

Chapter 2: Literature Review

Balance Control in Quiet Standing

An important relationship to discuss regarding stability during quiet standing control strategies is the relationship between centre of pressure (COP) and centre of mass (COM) movements. COM is defined as a theoretical point in which the sum of the weighted position vectors of all components of the system is zero. For the human system, the COM is the weighted average position of all segments of the body, where each segment is weighted according to its mass proportion. Additionally, the COM is the point in which forces can be applied to create linear translations without inducing changes in angular velocity (i.e., angular accelerations), whereas, applied forces that are not directed at the COM impose additional angular acceleration. The COP is defined as a specific point at which the ground reaction force vector is applied. It marks a weighted average of all pressures of the object that is in contact with the ground – in human gait, such object in contact with the ground is typically the foot (single support) or feet (double support).

When standing stationary, the gravitational force acting at the COM must be met with an equal ground reaction in the opposite direction (Winter, 1995). However, small sagittal and frontal plane displacements of the COM, as a function of small internal and external perturbations, must be accompanied by COP displacements, which generate restabilizing moments about the COM to ensure it remains within the base of support. For example, if the COM shifts in the anterior direction, the COP must also shift in the same direction until the COP lies anteriorly to the COM, which creates an external restabilizing moment to displace the COM posteriorly. COP displacements are predominantly products of internal joint moments of force created at the ankle (for A/P translations) and hip (for M/L translations). If the COM shifts to, or beyond, the limits of the BOS, subsequent shifts in COP may not be able to correct COM motion and therefore a change in support (i.e., stepping or grasping) may be needed to redefine the BOS to create more space for the COM to travel and create greater restabilizing moments of force about the COM (Winter et al., 1990).

Studies have gone on to map the COP-COM relationship and its importance as a stability control measure. Postural sway is a measure of COP excursions in relation to an individual's COM and is a common variable used to assess neuromuscular, sensory, and motor behavior associated with standing balance. Literature investigating the postural sway have suggested that as we age, the ability to modulate postural movements become more difficult; commonly measured by larger COP displacements during quiet stance (Muir et al., 2013; Weirich et al., 2010). Additionally, it has been suggested that increases in postural sway in older adult populations is linked to an increased risk of falls, however, it may offer insight on control strategies used to mitigate loss of balance (Muir et al., 2013). Early work addressing COP metrics and postural sway as a measure of stability suggested that older adults have similar COP displacements during quiet standing, however, significant group differences exist in COP displacements during voluntary forward and backward lean conditions (i.e., 50% lean and maximal lean conditions respectively) (van Wegen et al., 2002). Furthermore, it was found that older adults have decreased time-to-boundary metrics (i.e., the time needed for the COP to reach stability margins in the AP and ML directions) (van Wegen et al., 2002). Combined, these results suggest that increased postural sway seen in older adult populations may be indicative of decreased balance control. Further, decreases in spatial-temporal measures (i.e., COP time-to-boundaries) may be indicative of a sensory deficit in older adult populations. If older adults are unable to properly sense COM movements, decreased time-to-boundary may support the notion that older adults are producing excessive COP displacements (i.e., overshooting COM displacements in an effort to restrain COM motion before it crosses base of support limits and a subsequent step is required to regain balance. However, research has shown that older adults have decreases in distal muscle activations and increase ankle stiffness which may decrease generation of ankle plantar flexor moments (Tsai et al., 2013; Winter et al., 1998). This suggests that inability to generate enough force to efficiently arrest COM movements may be a result of motor challenges; but may also be indicative of sensory deficits if the generated moments are scaled inappropriately to actual instability yet scaled properly to perceived stability.

Feet In Place Control Strategies

Balance recovery can take one many forms, that is, different movement strategies can be executed depending on the external perturbation and the external environment. Feet in place movement strategies are characterized by constant BOS parameters however movement aims to quickly adjust COP location to arrest COM movement before whole body centre of mass crosses BOS limits. There are two common movement patterns seen during feet in place balance perturbations, that is, the hip and ankle strategies - both can be explained using the inverted pendulum model for balance recovery (Horak & Nashner, 1986; Winter et al., 1998). The ankle strategy utilizes plantar-flexors or dorsiflexors to shift COP in the anterior or posterior direction to shift the COM back towards an equilibrium state. The hip strategy is characterized using flexor or extensors about the hip to generate shear forces at the ground, taking up upper body momentum and shifting the COM towards the BOS. Authors found that individuals default to an ankle strategy when perturbed on a normal support surface, however, when the balance recovery task becomes more complex (i.e., shorter support surface), individuals adopt the hip strategy to recover balance (Horak & Nashner, 1986). Although these two strategies are only seen during feet in place balance preservation, this may indicate that less complex balance perturbation only require less complex movement strategies, that is, as magnitude of balance perturbation increases additional movement patterns may be utilized to regain a state of equilibrium (i.e., compensatory step).

Moreover, literature on different feet-in-place movement strategies has suggested that the choice in movement strategy is dependent on what the CNS is trying to control (Henry et al., 1998). It is suggested that a constraints model (movement is a product of task, environment, and goals) is a possible explanation for why the hip and ankle strategies are common feet in place strategies. Further, the CNS may prioritize kinematic features of balance control over other aspects present during balance recovery. Research has suggested that modulating COM movement through changes in COP are important mechanisms needed to regain balance and therefore the CNS may pick a movement strategy that favors an efficient way of moving the COM to an equilibrium state (Horak et al., 1997; Kuo, 1995; Winter, 1995). Authors have also postulated that the ankle and hip strategies are suited for different tasks, that is, the hip strategy is ideal for quickly moving the COM back towards equilibrium. This is due to the order of muscle activation which largely favors powerful hip flexors/extensors – potentially a movement strategy used when ankle strategy is no longer sufficient to arrest COM movement (Henry et al., 1995). Furthermore, the ankle strategy may be best suited for small balance perturbations that have minimal COM movements; this gives individuals the ability to keep the trunk in an erect position while adjusting the COM (Henry et al., 1995).

Age-related differences during fixed support balance control perturbations have also been noted in more recent literature. Brown et al. (1999) found that older adults and younger adults exhibit differences in desired control strategies when faced with balance perturbations to the support surface. It was found that across all trials younger adults use either the ankle or the hip strategy however, older adults are more likely to use a compensatory step. Further it was noted that older adults had a greater time-to-peak COM position which may suggest that older adults have trouble with sensing COM movement - therefore the COM can travel a greater distance before reactive movements occur. Moreover, older adults have decreased distance between the final COM positions in relation to the lateral aspect of the base of support following a compensatory step to regain balance (Brown et al., 1999). Brown et al. (1999) also used a cognitive task to better understand attentional demands. They found that when perturbed while doing a cognitive task older adults still favored the stepping response while younger adults showed little difference in recovery strategy. Such observed differences can however be explained by other literature that suggests older adults have less distal muscle activation, muscle power and torque resistance (Tsai et al., 2013), and the additional cognitive load of the secondary task may interfere with the temporal aspects of the feet-in-place response, which necessitates the more stepping response to safeguard stability. Additionally, is has been suggested that ankle, hip, and lumbar stiffness common among older adult populations may be contributing factors that inhibit older adults from adopting hip and ankle control strategies (Winter et al., 1998). Lastly, longer latencies during lower limb muscle activation have been reported in the older adult population (Woollacott et al., 1988). This may suggest that when older adults are perturbed their COM travels a greater distance before muscle activation begins, which may force the CNS to overcompensate and default to a compensatory step to regain balance.

Compensatory Stepping responses

Among compensatory steps in older and younger adult populations evidence shows that both groups use similar reactive control from step initiation up until foot contact. McIlroy &

Maki (1996) suggest that both groups share stepping characteristics, that is, swing durations, swing velocities, step lengths, and changes in centre of mass (COM) velocities during the initial step, along with the average times to onset of swing-leg unloading, to foot-off, and to foot contact. Although many features of compensatory steps are similar among age cohorts, research has shown that older adults have larger lateral COM motions which result in larger lateral step placement as a means to counteract this disturbance (Rogers et al., 2001). Further, literature has postulated that older adults exhibit challenges attenuating destabilizing limb accelerations produced by the perturbation; evident by larger momenta produced at the head and shoulder, which may reduce stability (Jensen et al., 2001). Lord & Fitzpatrick (2001) and Luchies et al. (2002) reported older adults having increases in reaction time prior to step initiation. This, coupled with lager COM deviations in older adult populations (Rogers et al., 2001), may be contributing factors when considering the challenges in minimizing destabilizing forces produced by instability. Although literature has shown some differences in spatial-temporal features of the stepping responses, there is little information that may explain why these challenges are apparent and what underlying mechanisms are responsible. Thus, research evaluating instability in different age groups shifts towards kinetic measures of instability during compensatory step reactions.

Considering the initiation and swing phases, work completed by Pijnappels et al. (2004) found that when taking a rapid step to recover from balance perturbations the push-off from the support limb provides additional time and clearance of the swing limb in the anteroposterior direction. This may aid in timing and positioning of the swing limb such that balance can be restored. Further, during push-off of the support limb, individuals were able to minimize or halt forward angular momentum of the body. This may suggest that during push-off of the support limb individuals orient the net ground reaction force (GRF_{net}) away from the whole-body COM in an attempt to arrest angular momentum or elevate force needed to regain balance subsequent to foot contact. Later work postulated older adults produce lower peak forces in both the AP and vertical direction during lower limb rapid stepping (Melzer et al., 2010). Further, older adults take longer to reach peak forces in the vertical direction during the preparation phase while exhibiting lower peak force generation. Such results may suggest that older adult populations struggled with quickly executing lower limb actions, and this may be the root of instability issues. Although, as prior research has shown, older adults are able to complete balance restoring steps to the same extent as younger adults (McIlroy & Maki, 1996); therefore, timing of the compensatory step may differ between age groups, but this may not be the root of instability related deficits. Melzer et al. (2010) also suggested that older adults may have larger center of mass displacements before achieving peak vertical force, suggesting that control of the COM movement during a balance restoring step may in part play a role in stability related challenges. This, coupled with greater instability and prevalence of multiple lateral steps following foot contact among older adults may suggest a scenario in which challenges arise after foot contact.

The restabilization phase focuses on movement subsequent to foot contact. Further, the restabilization phase plays a crucial role in the maintenance of stability, as the greatest opportunity to restabilise may occur in double-support, after initial limb contact. Regarding older adult populations the restabilization phase may prove as a viable way to extract mechanistic differences in the stepping responses following an unexpected balance perturbation. Such research in balance control focusing on the restabilization phase (subsequent to foot contact) has uncovered significant differences in latter responses to balance perturbations. Specifically, older adults exhibit challenges with mediolateral (ML) stability control following single step responses to balance perturbations. King et al. (2012) found that during compensatory stepping responses lateral kinematic features at foot contact are no different between younger and older adult participants; yet older adults show increases in mediolateral ground reaction and ankle forces. The authors suggest that these results support the idea that older adults maintain/regain stability by increasing force output, however, the analysis of reaction forces only considered those of the stepping limb rather than net forces. Later evidence suggests that older adults struggle with control of COM kinematics through the use of applied forces as a relationship between reactive control of frontal plane eccentricity of GRFnet to control lateral COM displacement (Singer et al., 2014, 2016). Further, examination of the restabilization phase of reactive stepping has shed light on two distinct time points. Early initial peak eccentricity of GRFnet (i.e., P1) that occurs within 100ms after foot contact is likely a result of limb stiffness prior to foot contact. P2 represents the second peak eccentricity occurring approximately 250ms after foot contact; taken together with prior research, this second peak may be a result of active force modulation to counteract instability during the restabilization phase (Henry et al., 1998; McIlroy & Maki, 1999; McIlroy & Maki, 1996). In comparison to younger adults, older adults produce similar P2 magnitudes but show increased time to P2 following foot contact. It remains unclear however, whether such increased time to the reactive restabilization response arises due to sensory challenges (detection of instability) or motor challenges (generating the kinetic response to instability) (Singer et al., 2016). Moreover, older adult populations exhibit measurable differences in lateral COM displacement relative to younger adult during the restabilization phase (Singer et al., 2016). Further, the measurable difference in COM displacement was positively correlated with the increase in P2 timing which may indicate that later applied forces during the restabilization phase might be a key feature in regulating balance control (Singer et al., 2016).

Relative to younger adults, older adults also exhibit challenges in the anteroposterior (AP) direction. Recent work examining age related differences in AP stability found that older adults continue to exhibit greater instability, as evidenced by kinematic measures of the COM (Singer et al., 2019). Additionally, challenges in applied force generation in relation to AP stability are different than that of ML stability, as older adults exhibited a decrease in P1 eccentricity magnitude and no age-related differences in P2 timing or magnitude; This may suggest that older adults struggle with proactive (rather than reactive) lower limb force generation in the AP direction. Furthermore, an inverse correlation between P1 magnitude and AP COM incongruity suggest AP instability may be due to challenges with applied force magnitude directly following foot-contact, suggesting AP stability may be controlled proactively and ML stability may be controlled reactively, after foot contact (Singer et al., 2019).

Components of The Gait Cycle

As steady state gait can be considered a series of controlled falls, it remains possible that previous measures used to uncover stability control mechanisms in perturbation-evoked stepping may be helpful in understanding balance control during steady state gait. Gait is a cyclical activity which consists of two phases. The stance phase marks the period in which the foot is in contact with the ground and makes of 60% of the total gait cycle. The swing phase represents the time in which the foot is off the ground moving toward a new point of contact; this period makes up the remaining 40% of the gait cycle. More recent work has noted that within the stance phase there are two independent tasks taken on by the limb, that is, weight acceptance and single support. Further, during the swing phase the tasks becomes limb advancement. Although the whole gait cycle is further broken down into eight additional stages within the three tasks, research has explained that humans are inherently unstable due to the COM traveling outside base of supports limits with most of the body weight is located in the head, arms, and trunk, at a considerable distance from the support surface (Winter, 1995As only one limb is in contact with the ground during the majority of the gait cycle (Winter, 1995), the double-limb support phase may be critically important to stability control, as there exists the possibility for interlimb control to regulate stability. However, during all periods of the stance phase there exists some potential to influence the trajectory of the whole-body COM through generation and modulation of lower limb force output and the consequent GRFnet.

The loading response phase occurs during the first 10% of the gait cycle and is characterized by the first period of double limb support. During the loading response phase, the stance foot is fully contacting the ground and body weight is transferred onto the stance limb. This phase is important for shock distribution, control of the vertical COM position, and forward progression (Kharb et al., 2011). Further, during this phase hip flexion is reduced, while the knee moves to a flexed position and ankle dorsiflexors control lowering of the foot to be fully in contact with the support surface (Kharb et al., 2011). The concentric extensor moment at the at the hip along with the eccentric extensor moment of the quadriceps about the knee control stability, absorb impact forces, and modulate knee flexion. Additionally, an eccentric plantarflexor moment is used to control the of dorsiflextion as the COM moves toward the stance limb (Winter, 1995).

The mid-stance phase occurs during 12% to 31% of the gait cycle and represents the time when the contralateral limb is no longer in contact with the ground and lasts until the body weight is aligned over the forefoot (Kharb et al., 2011). During mid-stance, the body moves forward over a stationary foot by ankle dorsiflexion, while the knee and hip are extended. Further, the mid-stance phase poses large challenges stability control systems as the base of support decreases significantly (i.e., one limb in contact with the support surface and the body's COM center of gravity moves to its highest and most lateral point through leg extension).

Terminal stance phase occurs as the heel of the stance limb rises and the limb continues over the forefoot, lasting until heel strike of the contralateral limb. This phase occurs during 31% to 50% of the gait cycle (Kharb et al., 2011). The knee increases its extension and then just begins to f1ex slightly. Increased hip extension puts the limb in a more trailing position. Terminal stance also poses challenges to stability, like mid-stance, motion is taking place on one limb and the center of gravity moves away from its highest point. Additionally, during this phase the heel strike of the contralateral limb becomes a concern for stability as the COM is redirected upward and laterally.

Lastly, the pre-swing phase occurs 50% to 60% of the gait cycle and represents the time after the swing leg contacts the ground. This phase begins when the prior swing limb contacts the ground and ends with stance limb toe off (Kharb et al., 2011). During this period, the body weight on the stance limb is unloaded and transferred onto the contra-lateral limb. The stance hip starts to return to 0 degrees from an extended position, while the knee starts to flex, and the ankle moves from dorsiflexion to plantar flexion (Kharb et al., 2011). During the pre-swing phase, the hip joint generates a concentric flexor moment that creates a propulsive force to aid in the initial swing during the swing phase. The knee joint continues to produce an eccentric extensor moment that controls knee flexion. At the ankle joint, there is a concentric plantar-flexor moment that aids in push-off the limb (Winter, 1995).

Kinetic features of the gait cycle for stability control

Ground reaction force (GRF) data have been extensively used to assess locomotion. The GRF can be broken into three sections, that is, point of application, magnitude and orientation. Point of application is said to move along the lateral portion of the foot during the stance phase (Winter, 1995), while orientations and magnitude fluctuate throughout the stance phase. In normal gait peak magnitudes of GRF during stance phase have been shown to be time locked at foot loading (~25% of stance) and push-off (~80% stance), both instances in which GRF exceeds body weight. Further, literature describing the orientation of GRF has suggested that unlike a compensatory step where individuals orientate GRF away from the COM to arrest angular movement, during locomotion the GRF vectors may be directed at or close to the COM in attempt to minimize angular momentum changes (Singer et al., 2014, 2016). Additionally, research has shown that the magnitude of vertical GRF components fluctuates with gait speed, and if directional demands are imposed, adaptations in horizontal GRF components are seen (Chiu & Wang, 2006; Dixon et al., 2014).

During the gait cycle sagittal, frontal, and transverse plane moments are produced for both propulsion and stability control. Joint moments during gait are greater in the direction of progression given that the goal is to move in the desired direction. During normal forward gait individuals are more stable due to larger MoS in the anteroposterior direction, however in the mediolateral direction individuals have a very small BOS and require more control of frontal plane moments to ensure stability throughout the gait cycle. As such the hip is the largest contributor to frontal plane stability; large frontal plane moments at the swing hip are used to control the trajectory of the swing limb while frontal moments of the stance hip are used to control the pelvis in relation to the upper body (Winter, 1995). In contrast, frontal plane moments generated at the knee and ankle are significantly smaller in magnitude. At the knee, abductor moments are used to stabilize the knee during the stance phase (a result of dominant adductor moments produced by the upper body); while the ankle produces small evertor moments during foot contact, invertor moments during mid-stance, and both invertor and evertor moments produced at the time of propulsion – likely used to aid in limb trajectory during swing (Eng & Winter, 1995).

Stability during Normal gait

Early researchers postulated that among younger and older adult populations there would be evident differences in stepping mechanics which may explain increases in fall risk. Double support time and step width were considered two indicators of fall risk, as they are linked to stability-related measures; individuals who adopt gait patterns with increased step width may be pre-emptively increasing BOS parameters to be ready for potential external forces causing instability. Further, increases in double support time would suggest that individuals have decreased their gait speed to better control subsequent steps; the double support phase also offers individuals a brief window to alter kinetic step mechanics to deal with external forces acting on the body. Early research investigating these measures found that between older and younger adult populations there are no significant differences in stride width and double support time, suggesting that there is no age-related decline in stability control due to changes in support patterns (Gabell & Nayak, 1984). Further, the authors go on to suggest that if age related changes in stability control occur, they may be a result of changes in lower limb control strategies (i.e., segmental coordination, joint moments/powers, and muscle activation schemes), however it was noted that older adults that are considered fallers or more at risk of falls may have underlying neurological pathologies that contribute to fall risk. Further work completed by Shkuratova et al.

(2004) aimed to build on the work of Gabell & Nayak (1984) by exploring spatiotemporal adaptations between older and younger adults, that is, changes in step width and double support time but also adding stride length and cadence. The authors found that similar results during preferred forward walking - both older and younger adult participants exhibited no age-related differences in all examined variables (stride width, length, step cadence, and double support time). Additionally, they observed that during a figure eight walking task, age-related differences in double support time were apparent. This may suggest that older adult populations may only exert different movement adaptations when the system is challenged. Further the figure-eight task produced significant increases in within group variability in all dependent variables, further suggesting that when an individual’s stability is challenged movement patterns may become less consistent as the task produces more stress on the individuals control systems (Shkuratova et al., 2004). The aforementioned work may suggest that in order to investigate age related challenges in stability control the task must be found difficult such that it produces a change in movement patten. Further, the study of stability control should not be limited to spatiotemporal characteristics as they may be underlying mechanisms managing stability such changes in spatiotemporal parameters are not needed.

During bipedal walking, individuals are inherently unstable in the mediolateral direction due to the smaller BoS compared to sagittal plane measures and COM must be redirected side to side in each subsequent step, rather than simply vaulting over the support limb (Winter, 1995). This low mechanical stability, therefore, must be regulated during walking to help prevent destabilizing forces that result in falls. Early work on spatial-temporal features of instability observed that in older adult populations spatial-temporal step variability is common among fallers and non-fallers, most notably variability among stride time, length, and width which may be indicative of walking instability (Hausdorff et al., 2001; Maki, 1997). Although both studies were used to link the fear of falling to the changes in movement patterns of older adults, it does in fact offer insight into features associated with being less safe and possibly features that are apparent across age cohorts. Furthermore, later research has observed that low amounts of stride time variability and low amounts of variability in stride length are commonly considered to be suggestive of a safe walking pattern in both older and younger adult populations (Beauchet et al., 2005; Dubost et al., 2006). Because a large portion of the gait cycle is spent moving on one limb and requires a great deal of control to ensure stability, the CNS may regulate stride time to combat instability. Beauchet et al. (2009) similarly found that younger adults showed variability in step length, stride time, swing time and stance time, with the largest amount of variability seen in step width. Further, older adults exhibited age related increases in all spatial-temporal categories yet were subject to more variation in the presence of altering gait speed. Although younger adults are considered to more stable, observed variability across age groups may suggest there is an acceptable amount of variation that is in fact indicative of some level of active stability control when completing a task such as walking (Bauby & Kuo, 2000; Kuo, 1995; McGeer, 1990).

When addressing static mechanical stability of humans, in order to remain in a state of equilibrium the COM generally musty lie within the BOS (Winter, 1995). For dynamic tasks such as walking, however, early work completed by Hof et al. (2005) proposed the use of the extrapolated center of mass (XCoM) as a measure to address instantaneous stability, since the existence of the COM within the BOS by itself is insufficient for dynamic stability. The extrapolated COM (XCoM) is a measure that is based on the inverted pendulum model of walking, which addresses stability as the relationship between an individual’s COM and BOS parameters. The XCoM moves beyond the static COM-BOS relationship (as in quiet standing), as it incorporates both the position and velocity of the COM and can be used to observe a dynamic margin of stability (i.e., defined as the distance between the XCoM and the edge of the BOS) (Hof, 2008). The MoS allows for observations of control strategies by using foot placement to control stability, that is, alterations in step width and length, in conjunction with the COM position-velocity relationship, which are indicative of a stable system. Early work observing changes in spatiotemporal gait features and the relationship to MoS found that as gait was disturbed, and spatiotemporal features were altered, both younger and older adults maintained invariant mean mediolateral MoS measures (Bierbaum et al., 2010; MacLellan & Patla, 2006). Although this work only included the mean MoS across trials, this suggests that when gait is disturbed, the CNS uses a control strategy that favors mechanical stability. In addition, McAndrew Young & Dingwell (2012) observed that when voluntarily manipulating gait characteristics (i.e., step width) increases in step-width caused decreases in MoS in the AP direction but increases in the ML MoS. This may suggest that the CNS controls gait patterns to optimize MoS parameters in both planes of motion; however, when increasing step width individuals may take shorter more frequent steps to maintain the same velocity while sacrificing AP MoS. More recent research has begun using local dynamic stability (LDS) as an index for stability during walking. LDS defined as the maximum time-finite Lyapunov exponent which measures subjects' ability to attenuate to small internal and external perturbations (Dingwell et al., 2001). Moreover, research has shown that adult populations will decrease local dynamic stability and increase ML MoS to combat perturbations in gait (Hak et al., 2012). These findings agree with research on dynamic stability in individuals with lower limb amputations, a group of subjects that are more locally unstable and use larger margins of ML stability to maintain stability (Gates et al., 2013; Hof et al., 2007). These features may suggest that the CNS favors control strategies which preserve mechanical stability (preservation of the whole-body COM) by decreasing and local stability and adjusting spatiotemporal parameters to increase ML MoS.

Individual Limb (Split Belt) Perturbations in Gait

Human locomotion remains the most important activity of daily living and is required to be flexible such that one can adapt to task and environmental constraints placed on the individual. Early work studying how adults adapt gait in the presence of novel gait patterns found that when the asymmetrical pattern is delivered, changes in single limb support time and stride time change immediately, along with changes in step length (Reisman et al., 2005). Additionally, authors found that there were no significant differences in joint angles between fast and slow legs (Reisman et al., 2005). This may suggest the CNS favors control strategies that offer less joint angle variability while changing spatiotemporal step parameters. Later work examining the ability to adapt gait parameters in individuals with neurological pathologies (i.e., Parkinson's disease and stroke) found that when using a split-belt paradigm participants were able to quickly adapt to situational demands (Reisman et al., 2007; Seuthe et al., 2020). It was found that in relation to baseline measurements, when subjects were put through the adaptation phase (i.e., uncoupled belt speeds) decreases in single support time on the fast belt occurred (regardless of normal or compromised limb); additionally, increases in single support time in the slow belt occurred (Reisman et al., 2007; Seuthe et al., 2020). Lastly, post-stroke patients and normal participants have shown after-effects from the adaptation phase; less variability in double support time, step length, and increases in interlimb phasing (i.e., symmetry in limb angles) (Reisman et al., 2007). The above research suggests that regardless of pathological differences in populations, preservation of stability remains a priority and adaptations in gait can be elicited to ensure dynamic stability;

although this raises the question on the degree to which populations can adapt gait mechanics to changing environmental stimuli.

Recent work has aimed to build off work of Reisman et al. (2005) and determine if step asymmetry induced via split-belt paradigm changes knee joint contact force (Syrett et al., 2021). In relation to Reisman et al. (2005), participants showed similar initial step length asymmetries with diminishing magnitudes as the adaptation phase progressed (Syrett et al., 2021). However, Syrett et al. (2021) found that when comparing knee joint contact forces there were no significant differences between the slow belt during early/late and baseline/late adaptation phases. While differences were measured with the limb walking on the fast belt only between the baseline/late adaptation phase. These results suggest that the initial asymmetry may only alter spatiotemporal parameters but due to the consistency of the treadmill action no changes in muscle activation occur at foot contact, which has been linked to increases in contact forces. Additionally, a study has shown that those who can adapt quickly to asymmetrical gait patterns show the same amount of lower extremity effort at the ankle & knee (i.e., percentage of maximal voluntary contractions) (Roper et al., 2021). These results suggest that difference in the AP direction may be minimal due to the inherent consistencies in the forward walking task however difference in ML control may be more distinct.

Further work has gone on to address COM motion and velocity changes during asymmetrical walking patterns. Because the split-belt paradigm requires two independent belts to be set at two velocities the walking speed of the individual would be the average of the two belts. Further, during each single support phase the body will be traveling at different speeds therefore the COM velocity may be inconsistent during asymmetrical walking. Tesio et al. (2021) found that initial strides of the adaptation phase exhibited a mean COM velocity that was greater than the mean velocity of the treadmill belts; additionally, in LA the COM velocity decreased below the mean velocity of the treadmill belts. These results may suggest that instability may occur during the initial delivery of the asymmetrical belt speeds, and the later return to symmetrical belt speeds, as evidenced by the mismatches in COM velocity; however, as individuals habituate to the novel gait, the COM motion better matches the belt speed and therefore gait becomes more stable and controlled.

Most recent work has begun to investigate reactive control of dynamic stability when asymmetrical gait patterns are delivered. When studying younger adults, it was found that step length symmetry decreased and became more variable during initial delivery of asymmetrical gait, however, as the adaptation phase continued symmetry returned closer to baseline; no changes in step width were measured (Darter et al., 2018). Additionally, AP MoS initially increased compared to baseline during initial adaptation phase, while ML MoS slightly decreased during initial adaptation; both AP & ML MoS returned closer to baseline measures during later adaptation phase (Darter et al., 2018). Increases in AP MoS maybe be a product of the direction of the perturbation, that is, the asymmetrical pattern is delivered in the AP direction and therefore may be the cause of the increased AP MoS. However, initial decreases in ML MoS may support the notion that the imposed asymmetrical gait destabilizes individuals in the ML direction because more focus is put on preserving AP MoS. Buurke et al. (2018) found that initial ML MoS of individuals limb (i.e., fast and slow limb) both increased during the initial onset of asymmetrical gait and were able to return to near baseline measures by LA phase. Further, the ML MoS of the fast limb had a larger increase when compared to the slow limb, while correlations were present between MoS and changes in ML COP (Buurke et al., 2018). Although during double support phases ML MoS is sacrificed to preserve AP MoS, work completed by Buurke et al. (2018) may imply that ML MoS is actively controlled during the single support phase, seen by increases in ML MoS in both fast and slow limbs. Further, this work suggests that continual changes in ML COP are present up until toe-off and may aid in the trajectory of subsequent steps.

Previous work has also acknowledged changes in segmental limb coordination and is effects of whole-body angular momentum. Liu, Macedo, & Finley (2018) found that healthy individuals have no changes in whole-body angular momentum as a result of changes in asymmetrical step patterns. Additionally, work has suggested that intersegmental limb coordination is altered during asymmetric gait, and these changes are a result of a shift in lower limb momentum to reach a target symmetry; yet no changes in whole body angular momentum can be observed (Liu & Finley, 2020). This may suggest that the CNS manages whole-body angular momentum by allowing variability in step parameters (i.e., step length/step width) and between limb coordination (position at specific time series) (Liu et al., 2018; Liu & Finley, 2020).

Although the use of asymmetrical gait patterns as a mode to investigate active control of dynamic stability is a relatively new area, many studies involving younger adult populations (Buurke et al., 2018; Darter et al., 2018; Liu et al., 2018; Liu & Finley, 2020; Tesio et al., 2021), do not include kinetic measures therefore there is precedent to examine kinetic differences in active control of dynamic stability in younger adult populations to understand the mechanisms responsible for stability control during asymmetrical gait perturbations. That is, changes in spatiotemporal features may be a primary response to instability, however, changes in magnitude and timing of applied forces to aid in offsetting angular accelerations about the COM may be a secondary response to stability control. Additionally, differences in applied forces between limb measures may give insight into understanding the implication of strategic variation (i.e., healthy CNS promote variation) between limbs to maintain whole-body stability. Further, this line of inquiry will aid in later comparisons across different populations to probe at age related differences in stability control.

Continuous Platform (Support Surface) Perturbations During Gait

Later work building off these findings sought to test how continuous platform oscillations (i.e., AP & ML translation via embedded instrumented treadmill capable of motion with six-degrees-of-freedom) would affect how individuals adapt their gait pattern to remain in a stable position (McAndrew et al., 2010). The authors found that when younger adults experience continuous pseudo-random perturbations in the AP direction, they exhibited decreases in step length while increasing step width when compared to no perturbation conditions (McAndrew et al., 2010). Further, when perturbed with ML translations participants elicited greater decreases in step length and increases in step width when compared to perturbations in the AP direction (McAndrew et al., 2010). Additionally, participants showed overall increases in step placement variability, with greatest measured variability in the ML direction (McAndrew et al., 2010). These results suggest that that individuals may struggle with regulating gait kinematics when mediolateral stability is challenged. Additional work was completed to assess stability measures, that is using floquet multipliers (FM) to measure orbital stability and local divergence exponents (LDE) to measure local dynamic instability (McAndrew et al., 2010). Using the same methodology, the authors found that younger adults are were able to maintain orbital stability ($maxFM<1$) and local instability ($\lambda^*>0$) across all trials (no perturbation, ML, and AP perturbations), with the greatest amount of movement variability during the ML perturbations (McAndrew et al., 2010). These results are constant with previous work that suggests instability in the ML direction is more taxing seen by greater amounts of movement variability when compared to AP and no perturbations trials (McAndrew et al., 2010). Furthermore, the above studies are consistent with biomechanical modeling studies that predicted that humans are more unstable in the ML direction compared to the AP (Bauby & Kuo, 2000).

More recent work evaluating gait adaptations in the presence of continuous perturbations sought to advance information surrounding dynamic stability control using MoS measures (Hak et al., 2012). Authors used the same methodology as previously mentioned studies (McAndrew et al., 2010, 2011) with the addition of MoS measures and found that younger adults exhibited very similar results. Similar to previous work, Hak et al. (2012) found that younger adults when faced with platform translations in the ML and AP direction, responded by decreasing step length, increasing step width, and increasing step frequency; but no reduction in speed was noted (Hak et al., 2012). Furthermore, spatial features of gait increased linearly as translation intensity increased. In regard to MoS, younger adults exhibited increases in posterior and mediolateral MoS as perturbation intensity increased (Hak et al., 2012). These results suggest that when gait is challenged individuals pick a control strategy which aims to increase the MoS, through modification of step parameters such as frequency, width, and length rather than by decreasing waking speed (Hof et al., 2007).

Differences Between Treadmill and Overground Walking

The use of treadmills in gait research has been increasing, as they allow for continuous monitoring of kinematic and kinetic parameters. Additionally, treadmill walking is stationary and can allow for longer walking sessions that may otherwise require large walkways to complete. However, numerous studies have questioned the efficacy of instrumented treadmills and the potential for differences in movement patterns that may not be generalizable to normal overground walking conditions. Early work addressing differences between the two modes of gait research found that among treadmill and overground walking there were subtle temporal and spatial differences in stepping parameters (Stolze et al., 1997). The research suggested that among healthy adults decreases in stride length, stance phase, double support time, and total cycle duration during are evident in treadmill walking; with increases being seen in cadence, step width, and swing phase (Stolze et al., 1997). The authors attribute the difference between overground and treadmill walking to be a product of the changes in cadence, however it is mentioned that decreases in optical flow during treadmill walking may influence changes in motor patterns (Stolze et al., 1997). The differences in walking parameters between the treadmill and overground conditions may also be due to a lack of familiarity to the treadmill surface. Stolze et al. (1997) do not mention a familiarization period in which participants are allowed to become accustomed to the novel condition. This may however aid in the explanation of difference between conditions as the initial data collection may reflect participant habituating to the new walking environmental and therefore may not produce an accurate testing scenario.

Later work building off of Stolze et al. (1997) work addressing difference among treadmill and overground walking protocols found that treadmill walking can accurately replicate overground walking mechanics. Riley et al., (2006) investigated mechanics of gait in both treadmill and overground, focusing on sagittal plane kinematic and kinetic parameters. The authors similarly found subtle differences in spatiotemporal features, along with kinematic, and kinetic differences in movement patterns (Riley et al., 2006). However, Riley et al. (2006) incorporated a familiarity phase such that participants could become comfortable with the new walking condition (i.e., treadmill walking) and calculated coefficient of repeatability (CoR) values for overground trial. Comparing CoR values of overground walking trials to treadmill trials revealed that differences in walking parameters lied within Calculated CoR values, suggesting that any measured differences in walking conditions lie within an acceptable range of variability and therefore imply that treadmill walking may be an adequate mode for assessing gait mechanics. Additionally work completed by Lee & Hidler, (2008) further confirmed that any observed differences in in sagittal plane gait mechanics are still valuable as the differences are no more variable than those collected overground suggesting that treadmill walking protocols are a reliable way measure gait mechanics.

Moreover, more recent work has gone on to address the changes in step width first noted by Stolze et al. (1997). Rosenblatt & Grabiner (2010) investigated whether step width variability and frontal plane dynamic stability (i.e., MoS) were altered when walking on a treadmill. The authors found that although step width increases during treadmill walking, variability during locomotion was found to decrease (Rosenblatt & Grabiner, 2010). Additional MoS measures were found to be invariant between walking conditions (Rosenblatt & Grabiner, 2010). The authors attribute the observed results to reflect that in new situational demands foot placement is chosen to maintain specific MoS (Rosenblatt & Grabiner, 2010). However, this may be the case when walking demands are challenged, but Rosenblatt & Grabiner (2010) do however mention that the treadmill used did not have handrails and no familiarity period was given during the protocol. Therefore, one could speculate that the observed differences are due to new situational demands that were not screened out and may not reflect accurate differences between overground and treadmill walking.

Additionally, a new study has investigated weather patterns of coordination are altered when walking on a treadmill surface and whether coordination variability fluctuates (S.-L. Chiu et al., 2014). The authors used continuous relative phase profiles of the hip-knee and knee-ankle during both stance and swing phases. It was found that when walking at similar speeds joint coordination remained similar during both conditions, with differences in variability only found during stance phase (Chiu et al., 2014).

Differences In Optical Flow Between Treadmill and Overground Walking

During gait it has been proposed that steering is based on optical flow, that is, the pattern of visual motion at the eye (Gibson, 1974). Additionally, to reach a goal, individuals must keep the focus of expansion in the direction or travel (Gibson, 1974). Work addressing whether individuals use optical flow to walk toward a goal found that when little to no optical flow is available the individual relies more heavily on egocentric direction (i.e., the direction of an object relative to one's self) (Warren et al., 2001). As optical flow increases, individuals make less errors while walking, becoming more centered when traveling towards their target. The authors suggested that locomotion is governed by a combination of optical flow and egocentric direction; when little optical flow is available behavior favors egocentric direction, but as optical flow increases it increasingly adjusts behavior (Warren et al., 2001). Therefore, control of walking can fluctuate at the discretion of optical flow available in an individual's environment.

In the case of over ground vs. treadmill walking some of the observed differences in movement patterns may be a product of optical flow. When walking overground, one is moving towards a target, therefore, incoming information is continuously changing. However, when treadmill walking, individuals are stationary with little changes in optical flow. Therefore, based on pervious work by Warren et al. (2001), the decreasing in optical flow present while walking on a treadmill may play a role in differences in observed gait mechanics (Chiu et al., 2015; Lee & Hidler, 2008; Riley et al., 2006; Rosenblatt & Grabiner, 2010). Furthermore, this hypothesis is supported by Carollo & Matthews (2002) who suggest that during locomotion kinematics of gait seem to be the desired control strategy.

Rationale

Walking remains one of the most important activities of daily living. Although steady-state gait involves fairly stable phase-locked relationships (i.e., 180 degrees out of phase) between limbs, even the act of navigating non-complex environments can necessitate transient modifications to the phase relationship between limbs, which may result in instability. While we understand the initial and adaptive temporospatial and kinematic gait alterations to split-belt gait perturbations among both younger and older adult populations, we lack knowledge surrounding the kinetic adaptations that may be responsible for such changes. The double support phase of the gait cycle offers a small window in which individuals can utilize the contributions from both limbs to restabilize and reorientate external forces to ensure stability in subsequent steps. By examining the whole-body COM kinematics and the kinetic parameters influencing such kinematic alterations, it is hoped the following work will provide insight into how force-related mechanisms are modified during the double support phase to maintain stability along with the adaptive changes in balance control during prolonged bouts of instability. Investigating the kinetic underpinnings of mediolateral stability control during gait may provide future insight into increased instability and falls among older adults.

The current exploratory study examined young adults and their ability to adjust mediolateral control mechanisms in the presence of a continuous perturbation. The use of split-belt continuous gait perturbations enabled the investigation of active control of MoS and how individuals change stepping characteristics as they habituate to a continuous perturbation. Further, this work is important because it will aid in mapping normative gait adaptations among younger adults during continuous perturbations, as a baseline for later comparison to groups of older adults or those with specific control challenges.

Objective

The aim of the proposed study was to understand how a continuous bout of instability (i.e., an asymmetrical gait pattern) affects stability control mechanisms in the mediolateral direction. The objectives were to examine mediolateral instability during asymmetrical walking using kinematic and temporospatial parameters (i.e., step length, width and frontal and sagittal plane MoS); determine force-related measures of instability during gait (i.e., angle of divergence, as a measure of the eccentricity of the net ground reaction force, which influences whole-body angular momentum); and examine timing and magnitude of potential proactive and reactive control indices of stability (i.e., initial (P1) and later phase (P2) net ground reaction force eccentricities) to better understand how individuals modulate forces in relation to situational demands to maintain stability.

Hypothesis

Early Adaptation Phase (EA)

Temporospatial Variables

In accordance with previous studies completed by Buurke et al. (2018), I hypothesised that during the initial delivery of an asynchronous gait pattern, fast limb step width and step width variability will increase relative to baseline.

Kinematic Variables

During the initial delivery I hypothesised that fast limb ML MoS will increase (i.e., more stable) and there will be increased step-to-step variability in the fast limb ML

MoS compared to baseline measures, likely due to fluctuations in ML temporospatial parameters and mismatches between COM kinematics and step parameters (Buurke et al., 2018).

Kinetic Variables

In response to initial delivery, I hypothesised that individuals would exhibit changes in some GRFnet orientation variables during double support. Specifically, I would expect to observe no change in the fast limb magnitude of P1 coupled with increases in magnitude of P2, relative to baseline, suggesting an increased need to utilize reactive force control to offset instability. Additionally, participants will demonstrate greater variability in both fast limb P1 and P2 magnitude as a result of initial changes in COM kinematics and increases in step width, suggesting challenges in precisely scaling force output to match changes in COM kinematics. It was believed that participants would exhibit similar time to P1 while increased time to P2 should be observed during the initial delivery, compared to baseline, in the fast limb as individuals attempted to increase whole body angular momentum control through the P2 eccentricity. Lastly, I hypothesized we should observe increased step-to-step variability in timing of both P1 and P2 peak eccentricities in the fast limb. The above hypotheses would suggest that when stability is challenged individuals use a movement strategy that relies more on reactive stability control when proactive measures due not immediately adapt and are insufficient to maintain balance, due to the novelty of the gait pattern.

Late Adaptation Phase (LA)

Temporospatial Variables

To align with previously reported data, once participants have fully acclimated to the novel gait parameters, I hypothesised that we would observe no change in fast limb step width, a decrease in step width variability relative to early adaptation phase, and participants will exhibit elevated values relative to baseline (Buurke et al., 2018).

Kinematic Variables

Post habituation, I believe that participants will exhibit reductions in fast limb ML MoS magnitude and variability relative to the initial habituation phase. These variables should return to baseline values as previously reported by Buurke et al. (2018).

Kinetic Variables

Post habituation I hypothesised we will observe a greater fast limb P1 and smaller P2 magnitudes relative to habituation, suggesting improved proactive stability control and reduced necessity for reactive control. Relative to baseline, however, it is hypothesized that P2 magnitude will remain increased in the fast limb, suggesting that improvements in proactive stability control are imperfect, requiring reactive alterations in force parameters evidenced by the increased P2 magnitude. Additionally, I believe participants would exhibit reduced fast limb variability in both indices relative to habituation, with similar values to baseline.

Further, fast limb time to P1 will remain unchanged relative to initial delivery and will have similar values when compared to baseline. The fast limb timing of P2 will be reduced post habituation but will remain increased relative to baseline levels, to generate greater restabilizing angular impulse. Fast limb trial to trial variability in timing of both

P1 and P2 will return to baseline values, post-habituation. This would suggest that once participants are acclimated to a novel gait, control strategies are reverted more to that of a normal gait cycle – that is individuals rely mainly on proactive mechanisms (i.e., the P1 eccentricity, necessary to redirect the COM during double support) which are said to be modulated in part by temporospatial stepping parameters and, with less emphasis on reactive control measures, which are needed when faced with new situational demands.

Chapter 3: Methods

Participants

Twenty-eight healthy young adults with no self reported musculoskeletal, cardiovascular, or neurological conditions/injuries were recruited to participate in the study (age 21-34yr, height 1.57-1.99m, weight 59.8-113.4kg, 15 females/13 males). Participants had no prior experience walking on a split-belt treadmill while belt speeds differed. All participants gave written informed consent.

Sample Size/Selection

This study aimed to recruit a total of 28 participants, which was obtained using $\alpha = 0.05$, $\beta = 0.2$, $f = 0.28$, which was the smallest effect size observed across all outcome variables of pervious work examining the effect of step placement and trial repetition on the magnitude and timing of the GRFnet eccentricity during the restabilisation phase of compensatory stepping (Singer et al., 2012). The alpha value is the probability of a Type I error occurring in any hypothesis test (false positive) while the beta value is the probability of a Type II error occurring in any hypothesis test (false negative), while Cohen's f is one measure of the effect size when using statistical tests. Cohen's f measures the magnitude of the experimental effect, 0.4 is considered large, 0.25 is medium, and 0.1 is small (Cohen, 1988). As such, the larger the effect size, the greater difference in group means, which translates into a smaller estimated sample size. Lastly sample size estimates were calculated using GPower (Version 3.1.9.6).

Inclusion/Exclusion

Inclusion criteria for study participation required participants be within 18-35 years of age and be able to walk on a treadmill unassisted for up to 1 hour unassisted. Participants would be excluded from the study if any of the following exclusion criteria were met: lower body injury occurring within the last year, presence of any self-reported chronic lower body musculoskeletal challenges, the presence of any cardiovascular or neurological conditions or injuries.

Consent

Participants were required to give informed consent before study participation as per the University of Manitoba Research Ethics Board. Participants were apprised on the risks of participating in light of the COVID-19 pandemic.

Recruitment

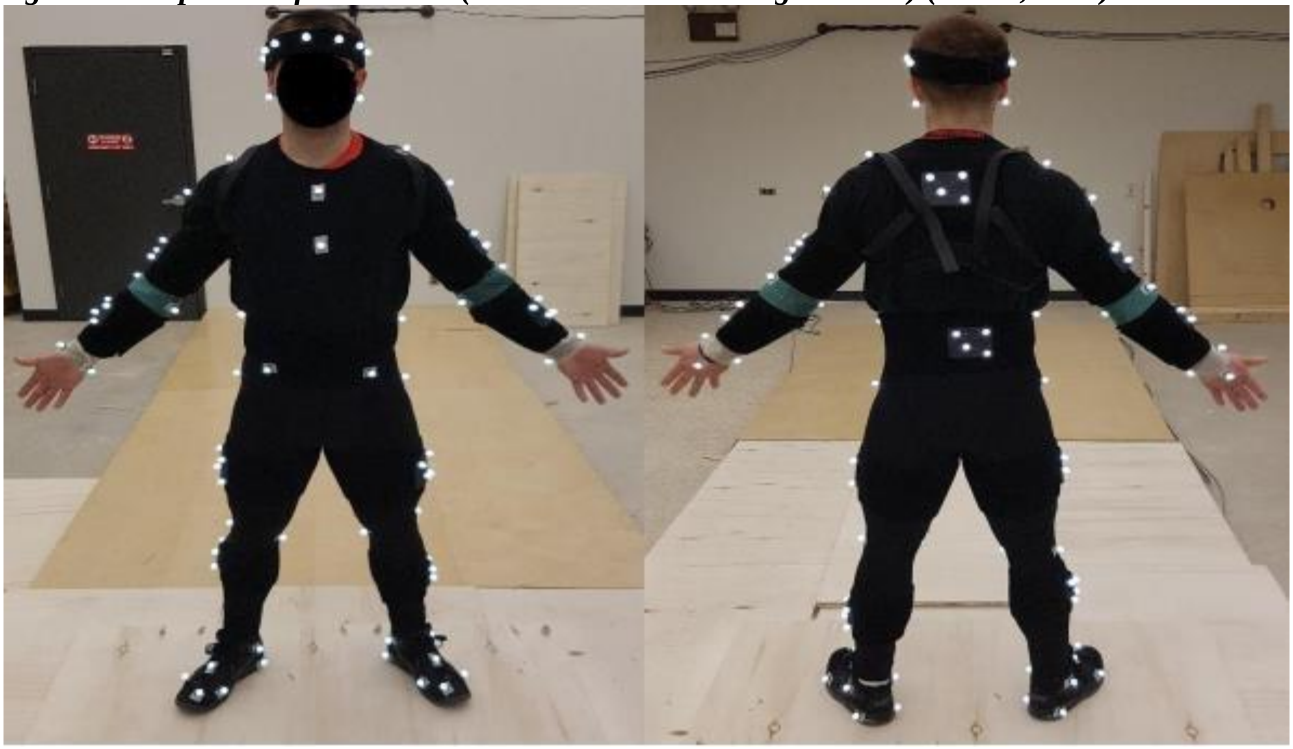
Emails were sent to various gyms and sporting centers (e.g., Sport Manitoba, Active Living Centre, etc.). Additionally, emails were distributed through the Faculty of Kinesiology and Recreation Management email list. Lastly, participants were recruited through word of mouth.

Instrumentation

Kinematic data was collected using 8 synchronized Vicon Vero 2.2 cameras (sampling frequency = 100 Hz) (Vicon Motion Systems, Los Angeles, CA). Kinematic data were used to model the position of the whole-body COM. Kinetic data (ground reaction force and moment data) were synchronously collected through the use of a Bertec split-belt instrumented treadmill (Bertec Corporation, Columbus, OH) which has two imbedded force platforms (sampling frequency = 1000 Hz).

Retroreflective markers were used to define anatomical landmarks on participants upper & lower limbs, and pelvis (Rawal, 2018; Singer et al., 2013). Calibration markers were placed bilaterally on the iliac crests, anterior superior iliac spines, acromioclavicular joints, anterior to the external auditory meatus, greater tubercles of the humerus, medial and lateral epicondyles of the humerus, radial and ulnar styloid processes, and the head of the 3rd metacarpal to define anatomical segments and provide limb metrics. Eight markers were placed on a headband to track position and orientation of the head. Four-marker tracking clusters were placed trunk and sacrum, along with bilaterally on the upper & lower arm, thighs, and shank. Calibration markers were used to locate anatomical structures, defining segment endpoints, which are used to construct a local coordinate system for each segment. After completion of marker calibration, calibration markers were removed. Rigid clusters of 4 tracking markers were placed on each segment – such marker clusters had fixed locations within each segment's local coordinate system and remained on the participant during experimental trials to track position and orientation of limb segments.

Figure 1 – Depiction of marker set (calibration and tracking markers) (Rawal, 2018).



Data Collection

A test for limb preference was required prior to study participation (Waterloo Footedness Questionnaire – see appendix A) to ensure recruited participants had a right foot preference (Elias et al., 1998). For this study both the left and right limbs were examined. Prior to data collection participants were required to complete a familiarization period on the split belt treadmill. The familiarization period had participants walk at two different gait speeds; the first was the slow period in which both belts will operate at 0.5m/s; participants would walk at this speed for 2 minutes. Participants would then walk at a fast speed, belts tied, at 1m/s for an additional 2 minutes. The familiarization periods helped participants habituate to treadmill gait and decrease the likelihood of missteps (i.e., stepping on the narrow surface that lies between the independent treadmill belts).

Participants would then complete two primary baseline phases, the first baseline in which the belts were coupled at 0.75m/s for 5 min, and the second baseline at 1.5m/s for 5min. Baseline measures were collected as an average of the last 10 strides during the baseline phase.

Upon completion of the baseline phase the treadmill would come to a complete stop; then a continuous treadmill perturbation was delivered. The right belt was set to 1.5 m/s while the left belt was set to 0.75m/s (i.e., 2:1 ratio). Participants walked continuously with the treadmill belts uncoupled for 15min. During this adaptation period specified kinematic and kinetic data were repeatedly monitored at 1 min intervals starting at the onset of the belt perturbation and terminating at the end of the 15-minute adaptation phase, to aid in observing habituation and determine at what point it occurred. The focus for analysis was placed on average values obtained from the first 10 strides following asymmetrical delivery and the last 10 strides of the adaptation block (i.e., last 10 strides of the 15th minute).

After the completion of the testing period participants would undergo a 5 min washout period in which the belts were re-coupled (set to 0.75m/s) to allow participants time to safely return to a normal gait patten. The participant was given the option to extend the washout period as needed to ensure they were comfortable with coupled walking upon leaving the laboratory setting.

Any missteps (i.e., foot placement that is not on treadmill belts) during data collection were noted. Additionally, missteps that occurred during the 10 strides required for outcome variable collection/analysis resulted in an extension of the trial length by an additional 10 strides (occurring a minimum of 10 strides after the misstep) to ensure accurate results. Further participants were allowed to extend the familiarization period (coupled belts) if needed to ensure they were comfortable going forward to the experimental trial. Participants were allowed to rest prior to perturbed walking if needed, however, once the perturbed walking had begun participants were no longer able to rest, without discontinuing the data collection. This was necessary to ensure that during this phase we could accurately collect data on biomechanical adaptations during perturbed walking. If a participant needed to stop the trial beyond this point, the participant’s data was removed from analysis.

Data Analysis

For this study kinematic and kinetic data were processed and analyzed using Visual 3D software (C- Motion, MD, USA Version 6) and Matlab (Mathworks, MA, USA Version 9.1 - R2021b), after individual marker trajectories and raw force plate data were extracted from Vicon nexus software. Individual marker data were low pass filtered using a zero-lag, twentieth-order critically damped filter, with a cut-off frequency of 6Hz (Robertson and Dowling, 2003). Force platform data were filtered using a low pass, zero-lag, fourth-order, Butterworth filter with a cut-off frequency of 15 Hz (Singer et al., 2014). The entire body was modelled as rigid system of individual segments, with 6 degrees of freedom. Segmental masses, centre of mass locations and moments of inertia for the present sample of younger adult participants were estimated using parameters provided by Zatsiorsky, Seluyanov, and Chugunova (1990) in conjunction with de Leva (2016), which included revisions to express data about segment endpoints, consistent with Dempster (1955). The whole-body centre of mass was calculated as a weighted average of each individual body segment, where each segment is weighted according to its mass proportion.

Temporospatial step parameters for each limb were calculated based on the individual limb calcaneus markers. Step length was calculated as the anteroposterior distance between individual limb calcaneus markers during double-support. Step width was measured as the ML distance between successive left and right heel-strikes. Variability of step length and width parameters were calculated based on step-to-step positional values, as the standard deviation of the 10 successive steps (each limb separately).

As a measure of frontal plane kinematic stability, ML-MoS was calculated - defined as the difference between an individual’s XCoM and the lateral edge of an individuals BOS (defined by the marker at the base of the 5th metatarsal) at any given instant in time. Current measures of dynamic margins of stability have been adapted from Hof et al. (2005). XCoM was calculated as:

$$XCoM = x + \frac{\dot{x}}{\omega_0}$$

where x is the COM-position, $\dot{x}$ is the horizontal COM velocity vector and

$$\omega_0 = \sqrt{g \backslash l}$$

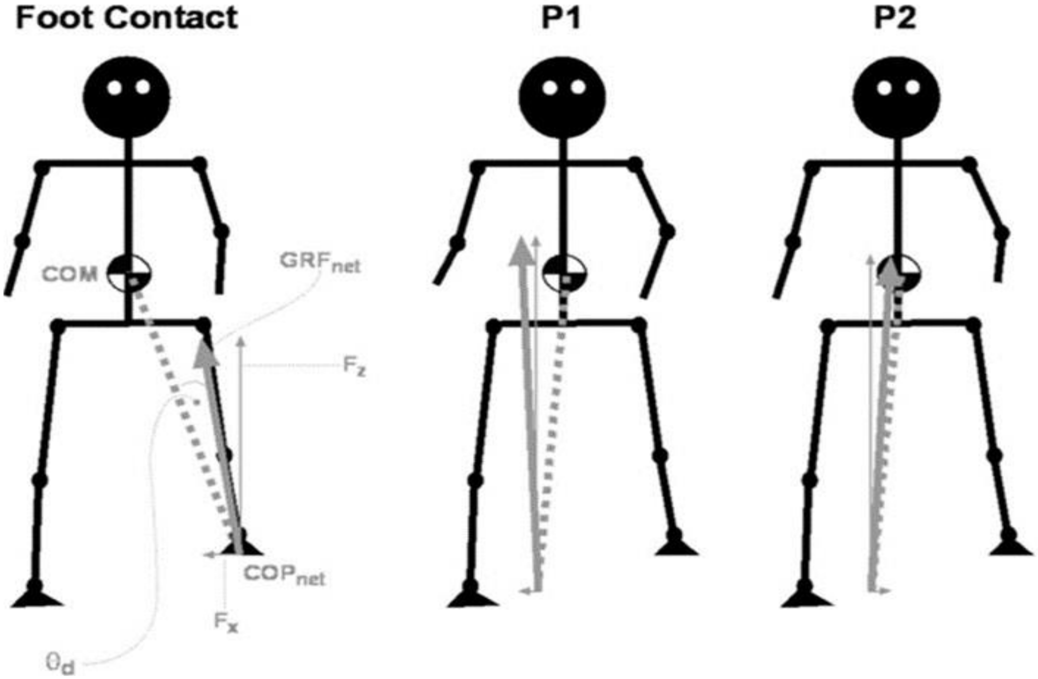
where g = 9.81m/s^2 is the gravitational constant and l is the pendulum length, that is, the mean difference from the lateral heel marker to COM. The MoS was defined as:

$$MOS = XCoM - BOS$$

where BOS was defined as the lateral boundary of the base of support (i.e., the marker on the base of the 5th metatarsal). Mediolateral MoS was calculated as the lateral distance between the XCoM and the lateral heel maker at foot contact. ML-MoS was calculated at every heel strike and at toe-off of the contralateral limb then averaged over 10 steps (for each limb). Step-to step variability of ML-MoS was defined as the within-participant standard deviation of the collected MoS data.

Custom code written in Visual 3D was used to calculate inclination angles of GRFnet within the frontal plane. Filtered force platform data from individual plates was combined to produce net ground reaction force and centre of pressure signals. Angles of inclination of the GRFnet and line joining the COP-COM was calculated relative to the mediolateral axis of the global coordinate system (i.e., the angle of each vector/line projected onto the frontal plane). Lastly, the COPnet is defined as the point of application of the net ground reaction force. The angle of divergence (θ_d) of the GRFnet was calculated as the difference between the frontal plane inclination angles of the GRFnet vector and COP-COM inclination angles. The angle of divergence represents the eccentricity of the net ground reaction force relative to the COM and acts to alter the linear and angular momentum of the body, through alteration of whole-body linear and angular accelerations.

Figure 2 - Angle of divergence (θ_d) between the net ground reaction force vector and of a line joining the net centre of pressure (COPnet) and whole-body centre of mass (COM) (Singer et al., 2016). The image depicts the ML angle of divergence for a right footstep.



Temporal measures of kinetic instability were also extracted in Visual 3D software. Initial and later phase positive peak θd (P1 & P2) have been adapted from Singer et al. (2016). For context, negative angle of divergence (θd) of the GRFnet relative to the COM at foot-contact indicates a destabilising effect of the GRFnet that results from the initiation and swing phases. The following two positive peaks (P1 and P2) represent the restabilizing effect of the GRFnet during the restabilisation (or double support) phase (of gait) (Rawal, 2021). The initial positive peak θd (P1) occurs within a 100ms immediately following foot-contact. The second positive peak θd (P2) occurs in a window 200-250ms following foot-contact. The magnitudes of the initial and later phase peak θd (P1 & P2) were calculated as the difference between GRFnet vector and the line connecting the COM and COPnet measured in degrees during the double support phase (i.e., when the respectively limbs are in the forward position).

Statistical Analysis

The aim of the proposed study is to investigate the kinetic and kinematic adaptations made during a continuous perturbation, with specific focus on the perturbed limb (i.e. on the fast belt) however, both limbs are analyzed for completeness. A one-way ANOVA model was used for statistical analysis. The experimental condition remained constant among all participants (0.75m/s slow belt, 1.5m/s fast belt). The dependent variables for this study were the kinematic variables (i.e., step width, step length, and ML MoS) and kinetic variables (i.e., ML angle of divergence (i.e., P1 & P2) timing and magnitudes) – all variables were subjected to the same ANOVA model.

For this study a two-factor (limb-time) within subjects repeated measures analysis of variance (ANOVA) was used. Whereby time had 3 levels (i.e., baseline, early adaptation, and late adaptation) and limb had 2 levels (left & right). Magnitudes (average of 10 steps within each condition) and intertrial variabilities (standard deviations of 10 steps within each condition) of each temporospatial and kinematic measure (step width, step length, and ML MoS) along with magnitudes, timings and intertrial variability of magnitude and timing of kinetic variables (angle of divergence (P1 & P2 eccentricities)) were subjected to the aforementioned ANOVA model. Mauchly’s test of sphericity was performed, if the assumption of Mauchly’s test was violated, the Greenhouse-Geisser test was used. Should a significant main effect of time be found, follow up paired t-tests were used to localize differences between the three time points. A Bonferroni correction was used to prevent alpha inflation as a result of the multiple comparisons. The alpha level for the omnibus ANOVA was set a $p<0.05$; for follow-up comparisons for a main effect of time, $p<0.017$ (i.e. 0.05/3). All statistical analyses were completed in SPSS (V. 28).

Chapter 4: Results

Twenty-eight participants were recruited for this study; all participants completed the protocol to its entirety. No participants fell during the experimental session. All participants completed a footedness questionnaire to ensure right leg dominance was constant across all individuals. No missteps (i.e., foot placement that is not on treadmill belts) were noted during data collection. Further, all participants stated that they were comfortable to complete the experimental trial; no participant's data was removed due to an incomplete experimental trial.

Below results are based off two baseline (BL) trials. During the slow limb baseline treadmill belts were tied at a velocity of 0.75 m/s and baseline data were extracted for the left limb. The fast limb baseline belt velocity was tied at 1.5 m/s and baseline data were extracted for the right limb. Data were extracted as an average of the last 10 steps of both baseline trials. Early adaption phase (EA) represents the first 10 steps following initial delivery of the split-belt perturbation, and later adaptation phase (LA) represents the last 10 steps of the fifteenth minute. Going forward the left limb is represented by slow limb data, while fast limb data represents the right limb.

Table 1 **– Participant characteristics**

	Male	**Female**	**Group Total**
N	13	15	28
Age	27(3.16)	26(2.72)	26(2.98)
Height (m)	1.82(0.07)	1.68(0.06)	1.74(0.10)
Weight (Kg)	83.86(15.7)	70.69(9.30)	76.80(14.11)

*Data presented as mean (SD).

Primary Analysis

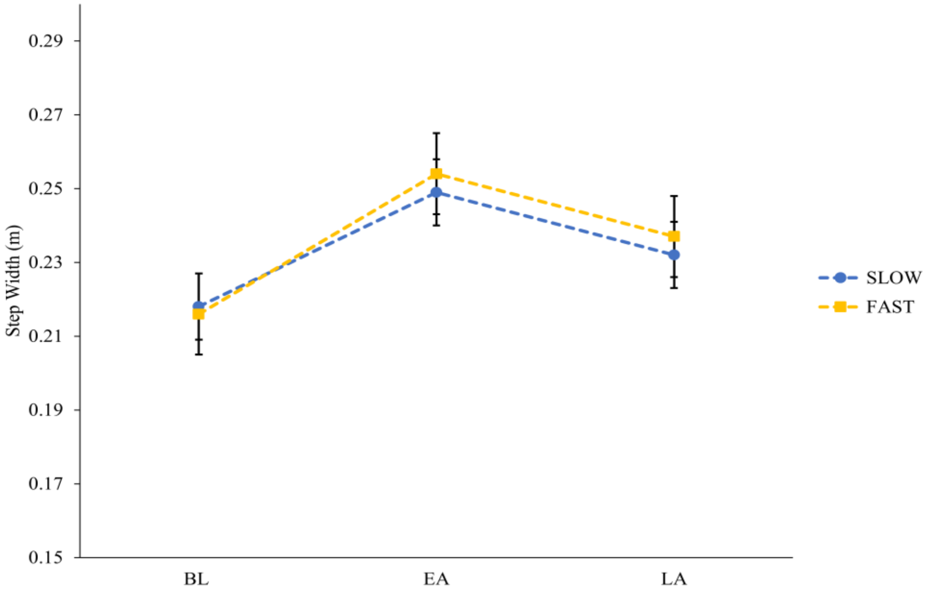
Temporospatial Variables

Step Width

The results showed that there was significant main effect of limb (F (1,27) = 16.252, p = <.001, $\eta p^2 = .376$); step width was less in the slow limb when compared to the fast limb. There was a main effect time (F (6.069,163.868) = 5.746, p = <.001, $\eta p^2 = .175$) whereby step width increased from BL at the onset of EA (p = <.001). Step width decreased between EA and LA (p = <.001), however, step width at LA was still greater than BL (p = .002). No limb by time interaction was observed.

Variability analysis revealed a main effect limb (F (1,27) = 11.860, p = .002, $\eta p^2 = .305$). The slow limb had less step width variability compared to the fast limb. There was a main effect time (F (8.205,221.526) = 15.086, p = <.001, $\eta p^2 = .358$) whereby step width variability increased when comparing BL to EA (p = <.001). Further, step width variability decreased when comparing EA to LA (p = <.001), however no difference was found when comparing BL to LA. No limb by time interaction was observed.

Figure 4 **– *Slow & Fast limb step width across BL, EA, and LA***



* Denotes a significance at p = <0.017

Table 2 **– Comparison of primary temporospatial variables between fast limb and slow limbs across all conditions**

	BL		EA		LA	
	SLOW	FAST	SLOW	FAST	SLOW	FAST
Step Width (m) †‡	.218(.043)	.216(.036)	.249(.035)	.254(.039)	.232(.033)	.237(.036)

*Data presented as mean (SD). † Denotes main effect time. ‡ Denotes main effect limb. § Denotes an interaction effect.

Table 3 **– Comparison of primary temporospatial variabilities between fast limb and slow limbs across all conditions**

	BL		EA		LA	
	SLOW	FAST	SLOW	FAST	SLOW	FAST
Step Width (m) †‡	.012(.004)	.017(.007)	.027(.011)	.028(.010)	.017(.008)	.017(.006)

*Data presented as mean (SD). † Denotes main effect time. ‡ Denotes main effect limb. § Denotes an interaction effect.

Kinematic Variables

Mediolateral (ML) Margin of Stability (MoS)

At The Instance of Heel Strike:

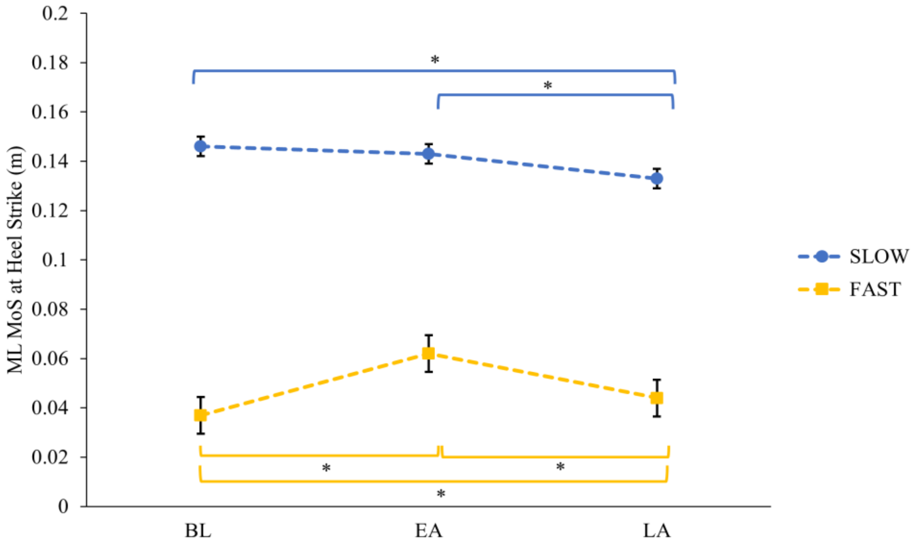
The results showed that there was significant main effect of limb (F (1,27) = 714.692, p = <.001, ηp^2 = .964) and a main effect time (F (5.122,138.299) = 18.481, p = <.001, ηp^2 = .406). There was also a limb by time interaction (F (4.743,128.059) = 21.795, p = <.001, ηp^2 = .447.

Results of follow up paired t-test revealed that slow limb ML MoS at heel strike was greater than fast limb at BL (p = <.001, d = 8.29), EA (p = <.001, d = 6.16) and LA (p = <.001, d = 5.74).

The results of the follow-up paired sample t-test revealed no change in ML MoS at the instance of slow limb heel strike at the onset of the EA when compared to BL, however ML MoS at the instance of fast limb (right) heel strike increased at the onset of EA when compared to BL (p = <.001, d = 1.67). ML MoS at heel strike decreased in both the slow and fast limbs when comparing LA to the EA (p = <.001, d = 6.16 and p = <.001, d = 1.16, respectively). Lastly, ML MoS at the instance of slow limb heel strike at LA was less than BL (p = .001, d = .99) and ML MoS at the instance of fast limb heel strike at LA was still greater than BL (p = .001, d = .45).

Variability analysis uncovered no main effect limb or limb by time interaction. However, a main effect time (F (5.991,161.747) = 28.200, p = <.001, ηp^2 = .511) was found, whereby variability of ML MoS at the instance of heel strike increased during EA when compared to BL (p = <.001). Variability decreased during LA when compared to EA (p = <.001), further no difference in variability was observed between LA and BL.

Figure 5 – Slow & Fast limb ML MoS at the instance of heel strike across BL, EA, and LA



* Denotes a significance at p = <0.017

Minimum Margins of Stability:

The results showed that there was significant main effect of limb (F (1,27) = 602.517, p = <.001, ηp^2 = .957) and a main effect time (F (5.342,144.243) = 16.335, p = <.001, ηp^2 = .377). There was also a limb by time interaction (F (5.042,136.129) = 17.975, p = <.001, ηp^2 = .400).

Results of follow up paired t-test revealed that slow limb minimum ML MoS following heel strike was greater than fast limb at BL (p = <.001, d = 7.70), EA (p = <.001, d = 6.98), and LA (p = <.001, d = 5.33).

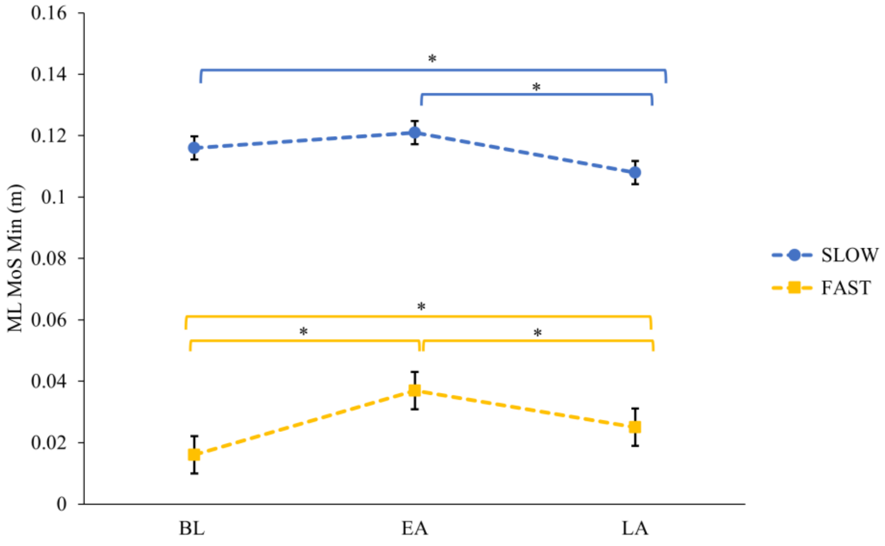
Slow limb minimum ML MoS exhibited no change while fast limb minimum ML MoS increased at EA when compared to BL (p = <.001, d = 1.44). Both slow and fast limb minimum ML MoS decreased when comparing LA to the EA (p = <.001, d = 1.03 and p = <.001, d = .79, respectively). Slow limb minimum ML MoS at LA was less than BL (p = <.001, d = .68) while fast limb minimum ML MoS at LA was greater than BL (p = <.001, d = .55).

Additional analysis on variability of minimum ML MoS revealed a main effect limb (F (1,27) = 13.958, p = .001, ηp^2 = .341), a main effect time (F (6.964,188.039) = 23.676, p = <.001, ηp^2 = .467), and a limb by time interaction (F (7.719,208.408) = 13.158, p = <.001, ηp^2 = .328).

Slow limb minimum ML MoS variability was less than fast limb at BL (p = <.001, d = .78), and at EA (p = <.001, d = 1.48). No significant differences in limb variability were observed during LA.

Follow-up paired t-tests found that the slow limb (p = <.001, d = 1.31) and fast limb (p = <.001, d = 2.23) increased at EA when compared to BL. Slow limb minimum ML MoS decreased at LA when compared to EA (p = .005, d = .66), while fast limb also decreased (p = <.001, d = 1.97). However slow limb variability was greater than BL when compared to LA (p = .008, d = .63), while no difference was observed in fast limb variability relative to BL.

Figure 6 – Slow & Fast limb minimum ML MoS after the instance of heel strike across BL, EA, and LA



* Denotes a significance at p = <0.017

Time from Heel strike To Minimum Margin of Stability:

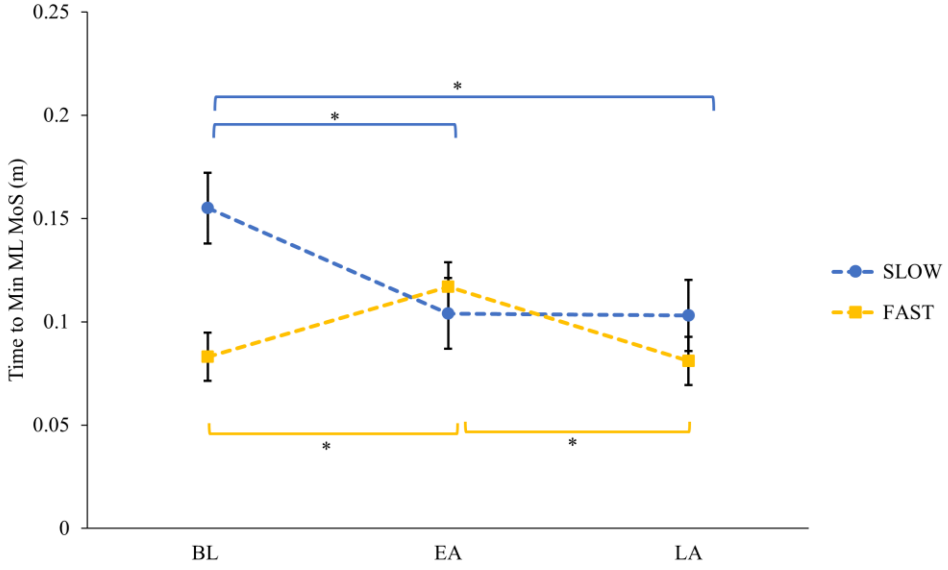
The results showed that there was significant main effect of limb (F (1,27) = 57.273, p = <.001, ηp^2 = .680) and a main effect time (F (3.850,103.959) = 44.537, p = <.001, ηp^2 = .623). There was also a limb by time interaction (F (3.599,97.168) = 43.071, p = <.001, ηp^2 = .615).

Results of follow up paired t-test revealed that slow limb time to minimum ML MoS following heel strike was greater than fast limb at BL (p = <.001, d = 3.34). No Difference was observed in time to minimum ML MoS following heel strike at EA. The slow limb has increased time to minimum ML MoS following heel strike when compared to the fast limb at LA (p = <.001, d = 1.44).

Additional follow-up paired sample t-test revealed that slow limb time to minimum ML MoS decreased (p = <.001, d = 1.92), while fast limb time to minimum ML MoS increased (p = <.001, d = 1.63) at the onset of the EA when compared to BL. Slow limb time to minimum ML MoS exhibited no change while fast limb decreased (p = <.001, d = 1.72) when comparing EA to LA. However, slow limb time to minimum ML MoS at LA was less than BL (p = <.001, d = 2.21) while fast limb was not significantly different than BL.

Additional analysis on variability of time to minimum ML MoS revealed a main effect limb (F (1,27) = 13.798, p = .001, ηp^2 = .338) whereby the slow limb has greater variability in minimum time to ML MoS than the fast limb. Further a main effect time (F (2.482,67.016) = 54.942, p = <.001, ηp^2 = .670) was observed whereby variability of time to minimum ML MoS increased (p = <.001) at EA when compared to BL. Further, variability decreased (p = <.001) when comparing LA to EA, however, no differences in variability were observed between LA and BL. No limb by time interaction was observed.

Figure 7 – Slow & Fast limb time to ML MoS following heel strike across BL, EA, and LA



* Denotes a significance at p = <0.017

Table 4 **– Comparison of mediolateral kinematic variables between fast limb and slow limbs across all conditions**

	BL		**EA**		**LA**	
	SLOW	**FAST**	**SLOW**	**FAST**	**SLOW**	**FAST**
ML MoS at Heel Strike (m) †‡§	.146(.011)	.037(.015)	.143(.011)	.062(.015)	.133(.015)	.044(.016)
ML MoS Minimum (m) †‡§	.116(.009)	.016(.016)	.121(.011)	.037(.013)	.108(.014)	.025(.017)
Time to Minimum Margins (s) †‡§	.155(.028)	.083(.012)	.104(.025)	.117(.027)	.103(.018)	.081(.012)

*Data presented as mean (SD). † Denotes main effect time. ‡ Denotes main effect limb. § Denotes an interaction effect.

***Table 5* – Comparison of mediolateral kinematic variabilities between fast limb and slow limbs across all conditions**

	BL		**EA**		**LA**	
	SLOW	**FAST**	**SLOW**	**FAST**	**SLOW**	**FAST**
ML MoS at Heel Strike (m) †	.006(.003)	.007(.003)	.018(.008)	.014(.004)	.007(.004)	.007(.004)
ML MoS Minimum (m) †‡§	.006(.002)	.008(.003)	.011(.005)	.020(.007)	.008(.004)	.008(.005)
Time to Minimum Margins (s) †‡	.020(.009)	.009(.009)	.063(.039)	.069(.040)	.019(.025)	.009(.003)

*Data presented as mean (SD). † Denotes main effect time. ‡ Denotes main effect limb. § Denotes an interaction effect.

Kinetic Variables

Mediolateral (ML) Angle of Divergence (θ_d) of the Net Ground Reaction Force

Initial Restabilisation Phase: (P1) Magnitude

The results showed that there was significant main effect of limb (F (1,27) =77.089, p = <.001, ηp^2 = .741) and a main effect time (F (6.399,172.770) = 5.208, p = <.001, ηp^2 = .162). There was also a limb by time interaction (F (4.802,129.652) = 22.645, p = <.001, ηp^2 = .456).

Results of follow up paired t-test revealed that slow limb P1 magnitude was less than fast limb P1 magnitude at BL (p = <.001, d = 4.28), EA (p = <.001, d = 1.13), and LA (p = <.001, d = 2.56).

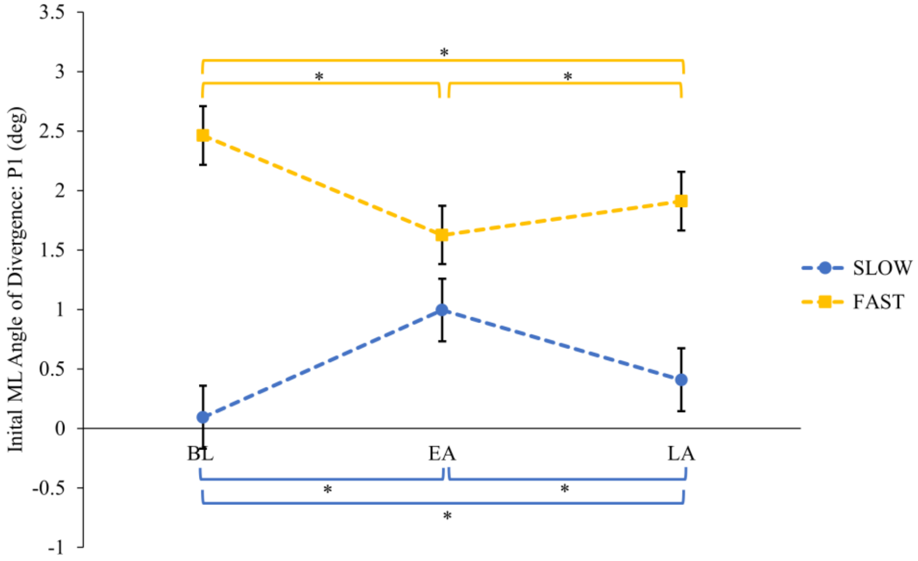
The results of the follow-up paired sample t-test revealed that slow limb increased (p = <.001, d = 1.56) while fast limb P1 decreased (p = <.001, d = 1.56) at EA when compared to BL. Slow limb P1 decreased (p = <.001, d = 1.00) when comparing LA to the EA whereas fast limb P1 increased (p = .004, d = .51). However, slow limb P1 was greater than (p = <.001, d = .60) BL when compared to LA while fast limb P1 at LA was less than BL (p = <.001, d = .91).

Variability analysis revealed a significant main effect of limb (F (1,27) =5.530, p = .026, ηp^2 = .170) and a main effect time (F (3.096,83.586) = 23.139, p = <.000, ηp^2 = .461). There was also a limb by time interaction (F (4.207,113.576) = 10.017, p = <.001, ηp^2 = .271).

Results of follow up paired t-test revealed that there was no difference between slow limb P1 and fast limb P1 variability at BL. At EA P1 magnitude was greater in the slow limb (p = <.001, d = .94) when compared to the fast limb. The slow limb had decreased P1 magnitude variability when compared to the fast limb at LA (p = <.001, d = .54).

The results of the follow-up paired sample t-test revealed that slow limb (p = <.001, d = 2.52) and fast limb P1 variability increased (p = .009, d = .74) at EA when compared to BL. Slow limb P1 variability decreased (p = <.001, 2.24) when comparing LA to the EA whereas fast limb P1 variability exhibited no change. However, slow limb P1 variability exhibited no difference when comparing LA to BL while fast limb P1 variability at LA was greater than BL (p = .012, d = .54).

Figure 8 – Slow & Fast limb P1 magnitude across BL, EA, and LA



*** Denotes a significance at p = <0.017**

Time To P1

The results showed that there was significant main effect of limb (F (1,27) = 11.510, p = .001, ηp^2 = .334) and a main effect time (F (1,27) = 11.510, p = <.001, ηp^2 = .157). There was also a limb by time interaction (F (5.243,141.552) = 6.907, p = .013, ηp^2 = .106).

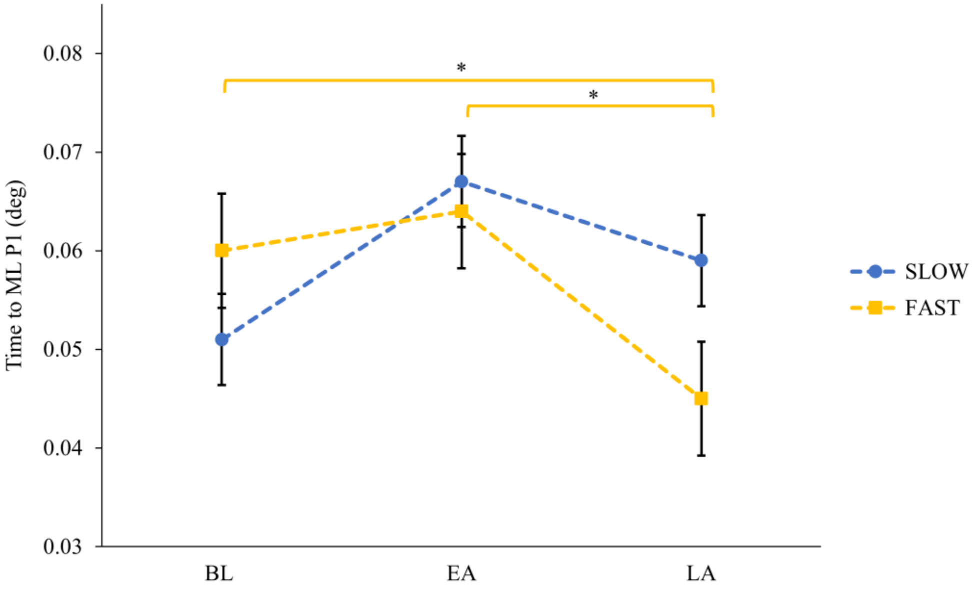
Follow-up paired t-tests revealed no difference between the slow and fast limb at BL and EA. However, slow limb time to P1 was greater than the fast limb at LA (p = .001, d = .86).

Additional paired t-tests revealed both slow limb and fast limb time to P1 exhibited no change when comparing EA to BL. There were no changes in slow limb time to P1 however fast limb time to P1 decreased (p = .001, d = .42) when comparing LA to EA. When comparing LA to BL, slow limb time to P1 exhibited no change while fast limb time to P1 decreased (p = <.001, d = .51).

Variability analysis revealed a main effect time (F (2.715,73.309) = 6.790, p = .001, ηp^2 = .201) whereby time to P1 variability increased at EA when compared to BL (p = <.001). During LA variability decreased when compared to EA (p = <.001), while no difference was observed when comparing LA to BL.

No main effect limb or limb by time interaction was observed.

Figure 9 – Slow & Fast limb time to P1 across BL, EA, and LA



* Denotes a significance at p = <0.017

Later Restabilisation Phase: (P2) Magnitude

The results showed that there was significant main effect of limb (F (1,27) =72.856, p = <.001, ηp^2 = .730) and a main effect time (F (4.815,130.013) = 9.708, p = <.001, ηp^2 = .264). There was also a limb by time interaction (F (2.563,122.994) = 3.926, p = .003, ηp^2 = .127).

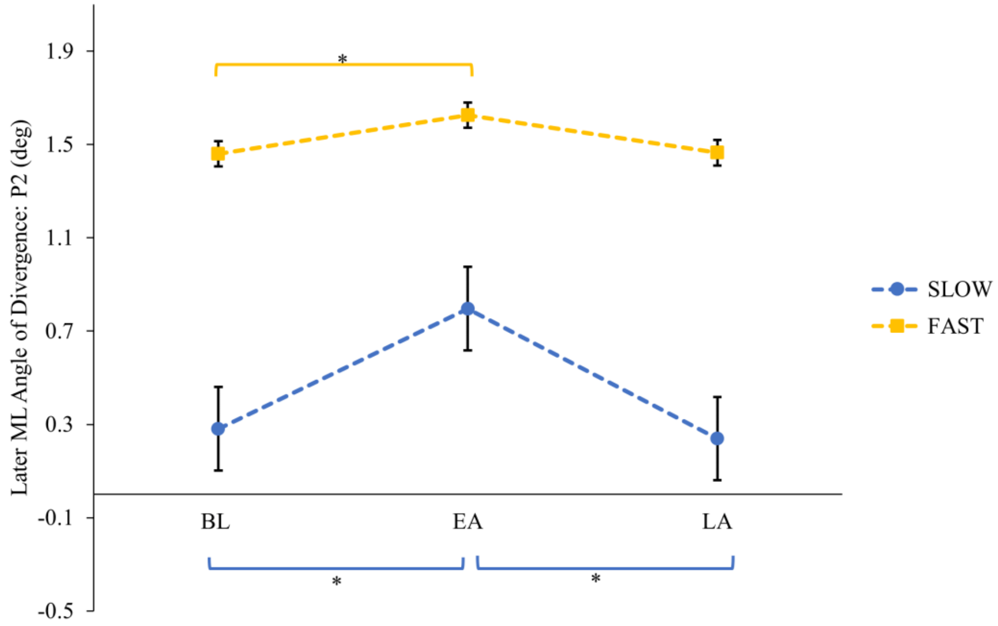
Results of follow up paired t-test revealed that slow limb P2 magnitude was less than fast limb P2 magnitude at BL (p = <.001, d = 2.12), EA (p = <.001, d = 1.20), and LA (p = <.001, d = 2.33).

The results of the follow-up paired sample t-test revealed that slow limb (p = .001, d = .78) and fast limb P2 (p = .008, d = .28) increased at EA when compared to BL. Slow limb P2 decreased (p = <.001, d = .83) when comparing LA to the EA whereas fast limb P2 exhibited no change. Both slow limb P2 and fast limb P2 exhibited no difference when comparing LA to BL.

Variability analysis revealed a significant main effect of time (F (3.392,105.658) = 16.316, p = .026, ηp^2 = .377) and a limb by time interaction (F (5.042,136.128) = 3.186, p = .009, ηp^2 = .106). No main effect limb was observed.

Follow-up t-tests revealed no significant differences between limbs across BL, EA, and LA. However, slow limb (p = <.001, d = 1.39) and fast limb (p = .002, d = .95) P2 variability increased at EA when compared to BL. Slow limb (p = <.001, d = .84) and fast limb (p = <.001, d = .1.10) P2 variability decreased during LA relative to EA. Lastly, P2 variability in both slow and fast limbs exhibited no differences when comparing LA to BL.

Figure 10 – Slow & Fast limb P2 magnitude across BL, EA, and LA



* Denotes a significance at p = <0.017

Time To P2

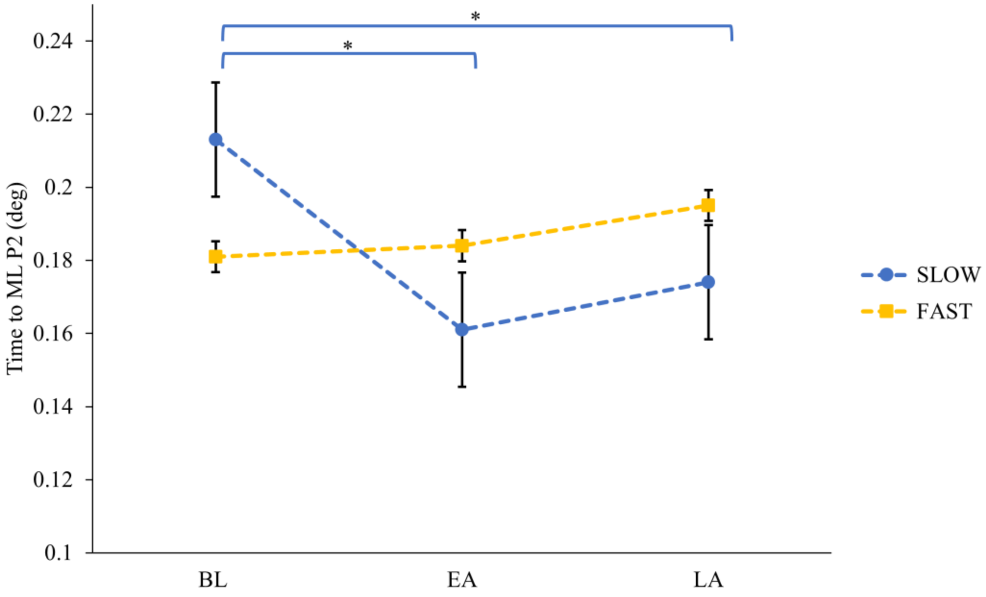
The results showed that there was significant main effect of limb (F (1,27) = 18.667, p = <.001, ηp^2 = .409) and a main effect time (F (5.072,136.955) = 7.202, p = <.001, ηp^2 = .211). There was also a limb by time interaction (F (6.427,173.419) = 12.272, p = <.001, ηp^2 = .312).

Results of follow up paired t-test revealed that slow limb time to P2 was greater than fast limb at BL (p = <.001, d = 1.20), however, slow limb was less than the fast limb at EA (p = <.001, d = .78). The slow limb had decreased time to P2 when compared to the fast limb at LA (p = <.001, d = .86).

Further follow-up paired sample t-tests revealed that slow limb time to P2 decreased at the onset of the EA when compared to BL (p = <.001, d = 1.87), while no difference in the fast limb time to P2 was observed. Slow and fast limb time to P2 exhibited no change when comparing EA to LA. However, slow limb time to P2 at LA was less than BL (p = <.001, d = 1.66) while there was no difference in right limb time to P2 at LA compared to BL.

Variability analysis revealed a main effect limb (F (1,27) = 15.263, p = .001, ηp^2 = .361) whereby the slow limb exhibited greater time to P2 variability than the fast limb. Further a main effect time was observed (F (5.066,136.792) = 21.995, p = <.001, ηp^2 = .449) whereby time to P2 variability increased at EA when compared to BL (p = <.001), decreased at LA when compared to EA (p = <.001), while no difference was observed at LA when compared to BL. No limb by time interaction was observed.

Figure 11 – Slow & Fast limb time to P2 across BL, EA, and LA



* Denotes a significance at p = <0.017

***Table 6* – Comparison of mediolateral kinetic variables between fast limb and slow limbs across all conditions**

	BL		EA		LA	
	SLOW	**FAST**	**SLOW**	**FAST**	**SLOW**	**FAST**
ML P1 (deg) †‡§	.094(.517)	2.462(.588)	.995(.629)	1.626(.478)	.409(.540)	1.910(.629)
Time to P1 (s) †‡§	.051(.031)	.060(.014)	.067(.022)	.064(.027)	.059(.021)	.045(.010)
ML P2 (deg) †‡§	.281(.531)	1.46(.579)	.796(.765)	1.626(.606)	.239(.561)	1.465(.489)
Time to P2 (s) †‡§	.213(.023)	.181(.030)	.161(.032)	.184(.027)	.174(.024)	.195(.025)

*Data presented as mean (SD). † Denotes main effect time. ‡ Denotes main effect limb. § Denotes an interaction effect.

***Table 7* – Comparison of mediolateral kinetic variabilities between fast limb and slow limbs across all conditions**

	BL		EA		LA	
	SLOW	**FAST**	**SLOW**	**FAST**	**SLOW**	**FAST**
ML P1 †‡	.249(.087)	.286(.116)	.834(.316)	.499(.391)	.294(.121)	.364(.129)
Time to ML P1 †	.009(.009)	.007(.005)	.030(.030)	.037(.037)	.012(.010)	.007(.004)
ML P2 †§	.228(.067)	.280(.140)	.605(.377)	.442(.196)	.286(.162)	.269(.106)
Time to ML P2 †‡	.019(.014)	.020(.013)	.027(.017)	.016(.008)	.028(.014)	.017(.011)

*Data presented as mean (SD). † Denotes main effect time. ‡ Denotes main effect limb. § Denotes an interaction effect.

Secondary Analysis

Spatiotemporal Variables

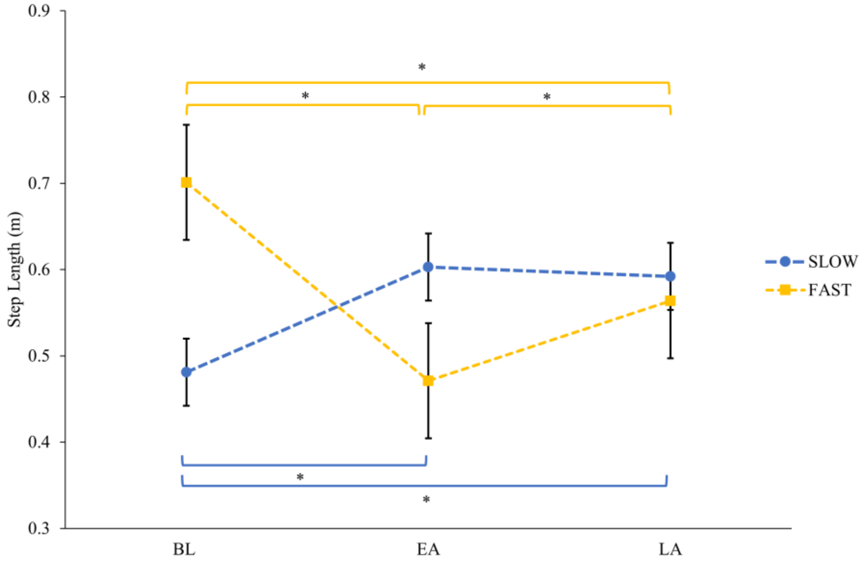
Step Length

The results showed that there was significant main effect of limb (F (1,27) = 45.061, p = <.001, ηp^2 = .625) and a main effect time (F (4.017,108.447) = 16.401, p = <.001, ηp^2 = .378). There was also a limb by time interaction (F (4.048,109.309) = 209.822, p = <.001, ηp^2 = .886).

Results of follow up paired t-test revealed that slow limb step length was less than fast limb at baseline (p = <.001, d = 6.96), however slow limb step length was greater than the fast limb at EA (p = <.001, d = 2.26) and LA (p = .001, d = .70). Follow-up tests on individual limbs show that slow limb step length increased (p = <.001, d = 2.78) while fast limb step length decreased (p = <.001, d = 4.61) from BL to EA. When comparing EA to LA the slow limb exhibited no change is step length, while the fast limb increased step length (p = <.001, d = 1.81). Both slow limb (p = <.001, d = 2.78) and fast limb (p = <.001, d = 4.34) step length did not return to BL measures.

Analysis of step-to-step variability revealed main effect limb (F (1,27) = 9.829, p = .004, ηp^2 = .267) in which the slow limb had greater step length variability than the fast limb. There was also a main effect time (F (15,405) = 2.564, p = .001, ηp^2 = .086), whereby variability increased from BL to EA (p = <.001), however decreased when comparing EA to LA (p = .013). There was no difference in step length variability when comparing LA to BL. No limb by time interaction was observed.

Figure 3 – Slow & Fast limb step length across BL, EA, and LA



* Denotes a significance at p = <0.017

Step Time

The results showed that there was significant main effect of limb (F (1,27) = 215.990, p = <.001, ηp^2 = .889) and a main effect time (F (3.565,96.259) = 5.037, p = .002, ηp^2 = .157). There was also a limb by time interaction (F (3.747,101.228) = 160.390, p = <.001, ηp^2 = .856).

Follow-up paired t-tests found that the slow limb had greater step time when compared to the fast limb at BL (p = <.001, d = 2.86). During EA the slow limb had decreased step time when compared to the fast limb (p = <.001, d = 1.18). Lastly during LA the slow limb had decreased step time when compared to the fast limb (p = <.001, d = 3.24).

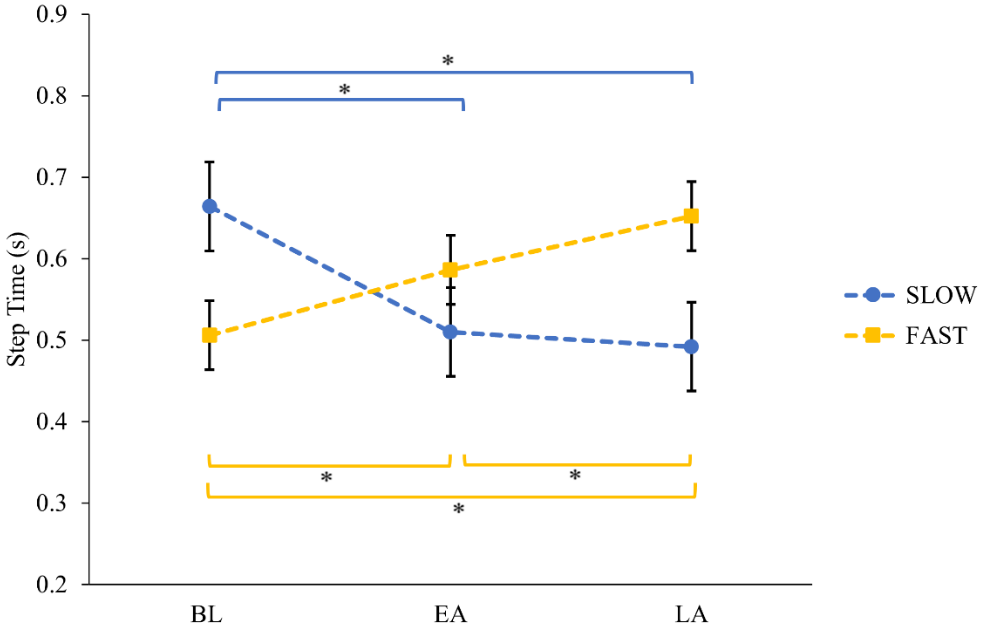
The results of the follow-up paired sample t-test revealed that slow limb step time decreased at the onset of the EA when compared to BL (p = <.001, d = 3.01) while fast limb step time increased (p = <.001, d = 1.18). Slow limb revealed no change at LA when compared to EA, however fast limb step time increased when comparing LA to the EA (p = .006, d = .98). Further, slow limb step time at LA was less than BL (p = <.001, d = 4.43) whereas fast limb step time at LA was greater than BL (p = <.001, d = 2.32).

Variability analysis showed that there was significant main effect of limb (F (1,27) = 28.623, p = <.001, $\eta p^2 = .515$) and a main effect time (F (1.507,40.702) = 127.860, p = <.001, $\eta p^2 = .826$). A significant limb by time interaction was observed (F (1.943,52.454) = 12.662, p = <.001, $\eta p^2 = .319$).

Follow-up paired t-tests found that the slow limb had greater step time variability when compared to the fast limb at BL (p = <.001, d = 1.23). During EA the slow limb had increased step time variability when compared to the fast limb (p = <.001, d = .58). Lastly during LA the slow limb had decreased step time variability when compared to the fast limb (p = <.001, d = .95).

Follow-up paired t-test revealed an increase in slow step time variability at the onset of EA when compared to BL (p = .001, d = 3.37) and fast limb step time increased variability (p = <.001, d = 2.54). Both slow (p = <.001, d = 3.55) and fast limb (p = <.001, d = 2.12) step time variability decreased at LA when compared to EA. Slow limb step time variability at LA was less than BL (p = .002, d = .82). Fast limb step time variability remained greater than BL (p = <.001, d = 1.17).

Figure 12 – Slow & Fast limb step time across BL, EA, and LA



* Denotes a significance at p = <0.017

Table 8 **– Comparison of secondary temporospatial variables between fast limb and slow limbs across all conditions**

	BL		EA		LA	
	SLOW	**FAST**	**SLOW**	**FAST**	**SLOW**	**FAST**
Step Length (m) †‡§	.481(.034)	.701(.029)	.603(.052)	.471(.064)	.592(.045)	.564(.034)
Step Time (s) †‡§	.664(.046)	.506(.063)	.510(.056)	.586(.072)	.492(.030)	.652(.063)

*Data presented as mean (SD). † Denotes main effect time. ‡ Denotes main effect limb. § Denotes an interaction effect.

Table 9 **– Comparison of secondary temporospatial variabilities between fast limb and slow limbs across all conditions**

	BL		EA		LA	
	SLOW	**FAST**	**SLOW**	**FAST**	**SLOW**	**FAST**
Step Length (m) †‡	.184(.006)	.014(.007)	.026(.010)	.024(.009)	.022(.008)	.018(.007)
Step Time (s) †‡§	.017(.007)	.010(.004)	.128 (.046)	.100(.050)	.012(.005)	.022(.014)

*Data presented as mean (SD). † Denotes main effect time. ‡ Denotes main effect limb. § Denotes an interaction effect.

Kinematic Variables

Anteroposterior (AP) Margin of Stability (MoS)

At The Instance of Heel Strike

The results showed that there was significant main effect of limb (F (1,27) = 167.351, p = <.001, ηp^2 = .861) and a main effect time (F (3.858,104.174) = 75.413, p = <.001, ηp^2 = .736). There was also a limb by time interaction (F (3.932,106.160) = 420.941, p = <.001, ηp^2 = .940).

Follow-up paired sample t-test revealed that slow limb AP MOS at the instance of heel strike was greater than the fast limb at BL (p = <.001, d = 5.90), however no difference was found during EA. Slow limb AP MOS at the instance of heel strike remained less than the fast limb at LA (p = <.001, d = 1.80).

The results of the follow-up paired sample t-test revealed that slow limb AP MOS at the instance of heel strike decreased (p = <.001, d = 5.10) whereas fast limb AP MOS at the instance of heel strike revealed no change at the onset of the EA when compared to BL. Slow limb AP MOS at the instance of heel strike at LA increased when compared to EA (p = <.001, d = 1.09) while fast limb AP MOS at the instance of heel strike at LA increased when compared to EA (p =

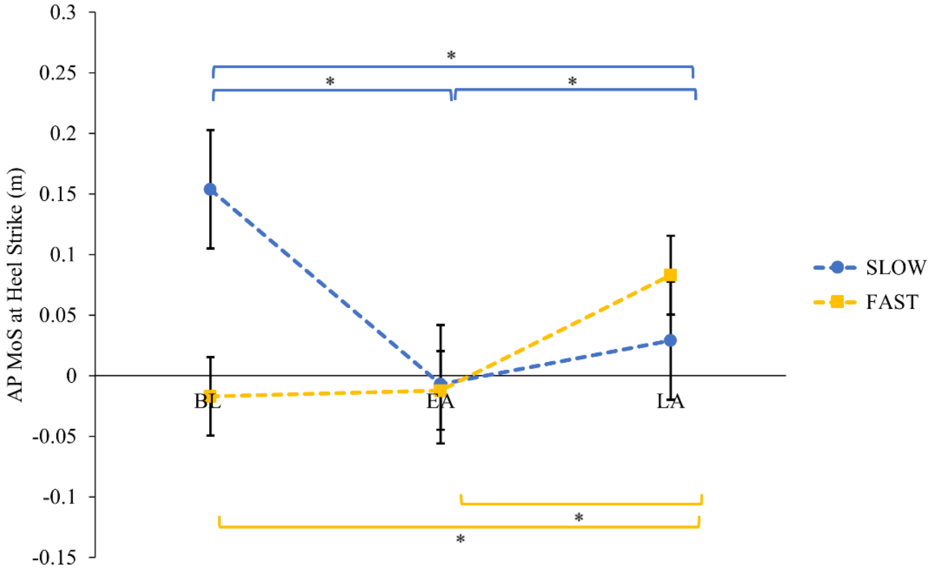
<.001, d = 2.53). However, slow limb AP MOS at the instance of heel strike at LA was less than BL (p = <.001, d = 4.09) while fast limb was greater than BL at LA (p = <.001, d = 3.51).

Variability analysis showed that there was significant main effect of limb (F (1,27) = 30.104, p = <.001, ηp^2 = .527) and a main effect time (F (1.972,53.250) = 48.465, p = <.001, ηp^2 = .642). A significant limb by time interaction was observed (F (2.021,54.568) = 7.627, p = .001, ηp^2 = .220).

Follow-up t-test analysis revealed that there was no difference between slow and fast limb AP MOS variability at the instance of heel strike at BL and EA, however slow limb AP MOS variability at the instance of heel strike was greater than the fast limb at LA (p = <.001, d = 1.00).

Additional variability analysis found that slow limb (p = <.001, d = 2.15) and fast limb (p = <.001, d = 1.83) AP MOS variability at the instance of heel strike increased when comparing BL to EA. Further, both slow limb (p = <.001, d = 1.79) and fast limb (p = <.001, d = 1.78) decreased between EA and LA. However slow limb variability at LA was greater than BL (p = <.001, d = .80), however, no significant difference was present in the fast limb.

Figure 13 – Slow & Fast AP MoS at the instance of heel strike across BL, EA, and LA



* Denotes a significance at p = <0.017

Minimum AP MoS:

The results showed that there was significant main effect of time (F (4.246,97.878) = 114.647, p = <.001, ηp^2 = .784). There was also a limb by time interaction (F (3.631,98.036) = 251.321, p = <.001, ηp^2 = .903). No main effect limb was observed.

Results of follow up paired t-test revealed that slow limb minimum AP MoS was greater than fast limb at BL (p = <.001, d = 7.71), EA (p = <.001, d = 2.55), and LA (p = <.001, d = .82).

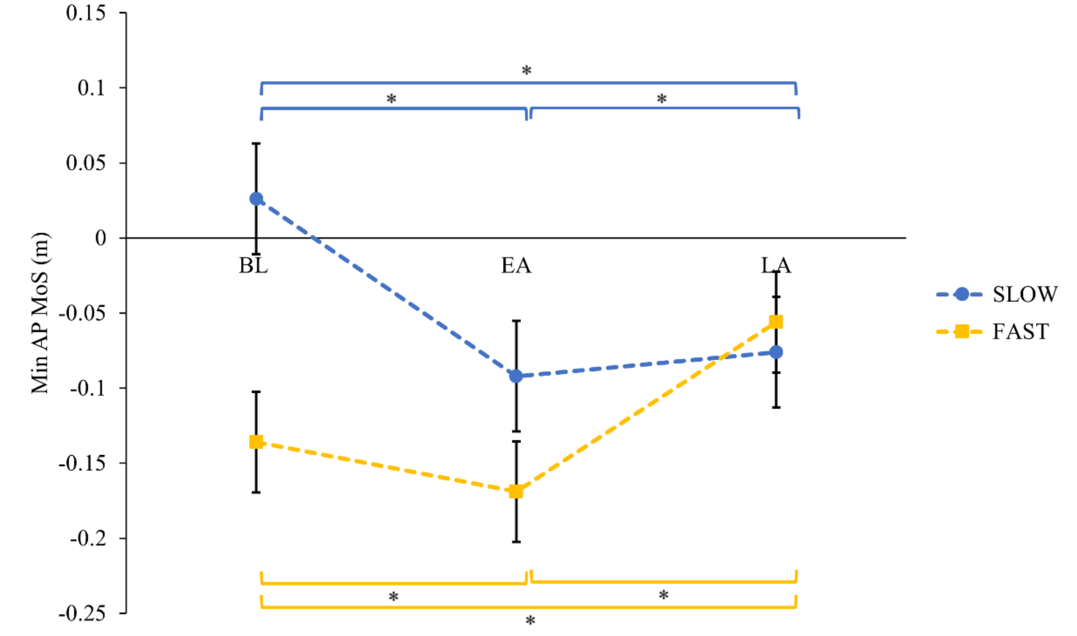
The results of the follow-up paired sample t-test revealed that slow limb minimum AP MoS decreased (p = <.001, d = 4.88) while fast limb AP MOS at the instance of toe off decreased (p = <.001, d = 1.19) at the onset of the EA when compared to BL. When comparing EA to LA, both slow limb (p = <.001, d = .61) and fast limb (p = <.001, d = 3.92) increased minimum AP MoS. Slow limb AP MOS at the instance of toe off at LA was less than BL (p = <.001, d = 4.42) while the fast limb was greater than BL (p = <.001, d = 3.55).

Variability analysis showed that there was significant main effect of limb (F (1,27) = 9.496, p = .005, $\eta p^2 = .260$) and a significant main effect time (F (2.949,79.632) = 121.662, p = <.001, $\eta p^2 = .818$). Additionally, a limb by time interaction was observed (F (3.007,81.177) = 3.418, p = .021, $\eta p^2 = .112$).

Follow-up paired t-tests found that the slow limb had less minimum AP MoS variability at BL (p = <.001, d = .47). Further no differences were found between limbs across time.

Additional follow-up paired t-tests revealed slow limb (p = <.001, d = 3.81) and fast limb (p = <.001, d = 3.73) minimum AP MOS variability at EA increased when compared to BL. Further, slow limb (p = <.001, d = 3.48) and fast limb (p = <.001, d = 3.71) minimum AP MOS variability at LA decreased when compared to EA. Lastly, slow limb minimum AP MOS variability was greater at LA than BL (p = .004, d = 1.00). Fast limb minimum AP MOS variability revealed no difference when comparing LA to BL.

Figure 14 – Slow & Fast AP MoS Minimum across BL, EA, and LA



* Denotes a significance at p = <0.017

Time From Heel strike to Minimum AP MOS

The results showed that there was significant main effect of limb (F (1,27) = 209.448, p = <.001, ηp^2 = .886) and a main effect time (F (3.394,91.640) = 9.086, p = <.001, ηp^2 = .252). There was also a limb by time interaction (F (3.856,104.117) = 23.054, p = <.001, ηp^2 = .461).

Follow-up paired sample t-test revealed that slow limb time to minimum AP MoS following heel strike was greater than the fast limb at BL (p = <.001, d = 5.34), EA (p = <.001, d = 2.26), and at LA (p = <.001, d = 2.44).

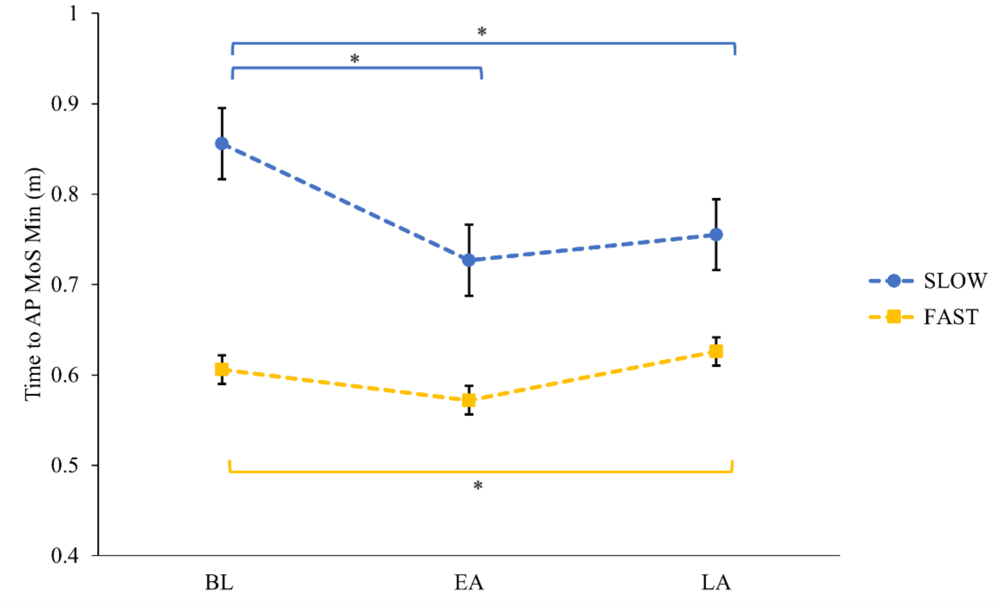
The results of the follow-up paired sample t-test revealed that slow limb time to minimum AP MOS at heel strike decreased (p = <.001, d = 1.80) at the onset of the EA when compared to BL, however the fast limb exhibited no change. No differences were observed in slow limb or fast limb time at LA when compared to EA. Slow limb time to minimum AP MOS following heel strike at LA was less than BL (p = <.001, d = 0.08), while fast limb time to minimum AP MOS following heel strike at LA was greater than BL (p = <.001, d = 1.25).

Variability analysis showed that there was significant main effect of limb (F (1,27) = 39.914, p = <.001, ηp^2 = .596), a main effect of time (F (1.252,33.815) = 124.657, p = <.001, ηp^2 = .822), and a limb by time interaction (F (1.865,50.368) = 12.876, p = <.001, ηp^2 = .323).

Follow-up analysis revealed the slow limb was greater than the fast limb at BL (p = <.001, d = 1.22). During EA, slow limb time to AP minimum MoS variability was less than fast limb variability (p = .004, d = 2.43). However, during LA slow limb variability was greater than fast limb variability (p = <.001, d = 0).

Additional follow-up tests revealed that both the slow (p = <.001, d = 2.18) and fast limb (p = <.001, d = .19) increase variability at EA when compared to BL. Further during LA variability decreases in slow limb (p = <.001, d = 2.16) and fast limb limbs (p = <.001, d = 1.27) when compared to EA, however, variability revealed no difference in either limb when comparing LA to BL.

***Figure 15** – Slow & Fast time from heel strike to minimum AP MoS across BL, EA, and LA*



* Denotes a significance at p = <0.017

Table 10 **– Comparison of anteroposterior kinematic variables between fast limb and slow limbs across all conditions**

	BL		EA		LA	
	SLOW	**FAST**	**SLOW**	**FAST**	**SLOW**	**FAST**
AP MoS at Heel Strike (m) †‡§	.154(.029)	-.017(.029)	-.007(.034)	-.012(.045)	.029(.032)	.083(.028)
AP MoS Minimum (m) †‡§	.026(.021)	-.136(.021)	-.092(.027)	-.169(.033)	-.076(.025)	-.056(.024)
Time to AP MoS Minimum (m) †‡§	.856(.059)	.606(.030)	.727(.083)	.572(.050)	.755(.065)	.626(.037)

*Data presented as mean (SD). † Denotes main effect time. ‡ Denotes main effect limb. § Denotes an interaction effect.

Table 11 **– Comparison of anteroposterior kinematic variabilities between fast limb and slow limbs across all conditions**

	BL		**EA**		**LA**	
	SLOW	**FAST**	**SLOW**	**FAST**	**SLOW**	**FAST**
AP MoS at Heel Strike (m) †‡§	.011(.005)	.009(.005)	.035(.015)	.047(.029)	.015(.005)	.010(.005)
Minimum AP MoS (m) †‡§	.010(.004)	.013(.008)	.057(.017)	.170(.059)	.014(.004)	.014(.008)
Time to AP MoS Minimum (m) ‡	.024(.012)	.012(.007)	.135(.071)	.013(.003)	.025(.013)	.025(.013)

*Data presented as mean (SD). † Denotes main effect time. ‡ Denotes main effect limb. § Denotes an interaction effect.

Kinetic Variables

Anteroposterior (AP) Angle of Divergence (θ_d) of the Net Ground Reaction Force

Initial Restabilisation Phase: (P1) Magnitude

The results showed that there was significant main effect of limb (F (1,27) = 281.160, p = <.001, ηp^2 = .912). There was also a limb by time interaction (F (5.558,150.076) = 106.146, p = <.001, ηp^2 = .797). No main effect time was observed.

Follow-up paired sample t-test revealed that slow limb AP P1 was less than the fast limb at BL (p = <.001, d = 3.05), however, greater than the fast limb at EA (p = <.001, d = 3.21). Slow limb AP P1 remained greater than the fast limb at LA (p = <.001, d = 2.83).

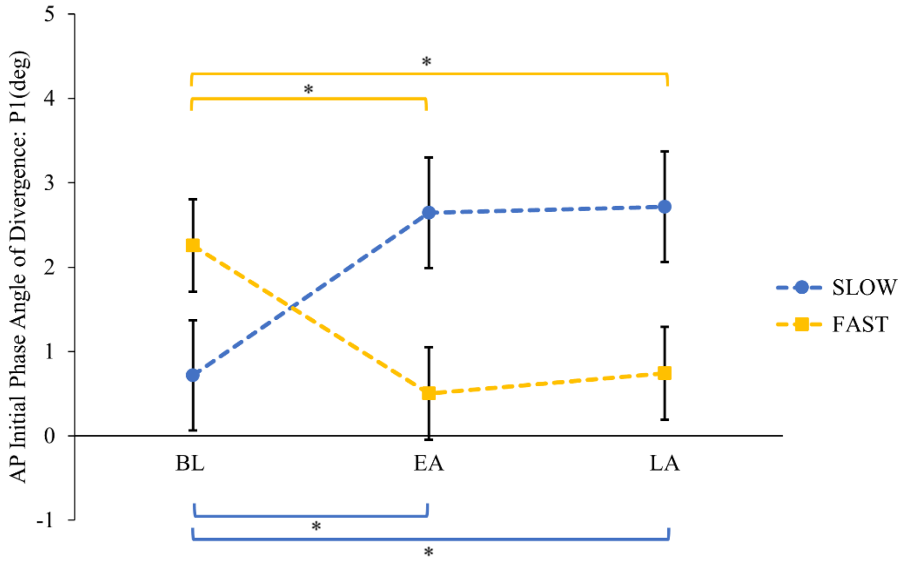
The results of the follow-up paired sample t-test revealed that slow limb P1 increased (p = <.001, d = 3.08) while fast limb P1 decreased (p = <.001, d = 3.17) at EA when compared to BL. No difference in slow limb and fast limb P1 was found at LA when compared to EA. Slow limb P1 at LA was greater than BL (p = <.001, d = 3.71) whereas fast limb P1 at LA was less than BL (p = <.001, d = 2.27).

Variability analysis showed that there was significant main effect of limb (F (1,27) = 15.695, p = <.001, ηp^2 = .368) and a main effect time (F (2.650,71.550) = 57.153, p = <.001, ηp^2 = .679). There was also a limb by time interaction (F (2.982,80.514) = 23.966, p = <.001, ηp^2 = .470).

Follow up t-tests revealed no difference in AP P1 variability at BL, however, at EA the slow limb has greater variability when compared to the fast limb (p = <.001, d = 1.66). At LA there was no significant difference between limbs.

Additional t-test revealed that both slow (p = <.001, d = 2.69) and fast limb (p = <.001, d = 2.09) AP P1 variability increased at EA when compared to BL. When comparing EA to LA, it was revealed that both slow (p = <.001, d = 2.41) and fast limbs (p = <.001, d = 1.38) decreased; further slow limb (p = <.001, d = 1.10) and fast limb (p = <.001, d = .60) variability was greater than BL at LA.

Figure 16 – Slow & Fast AP P1 magnitude across BL, EA, and LA



* Denotes a significance at p = <0.017

Time to P1

The results showed that there was significant main effect of limb (F (1,27) = 10.123, p = .004, ηp^2 = .273) and a main effect time (F (3.926,105.993) = 24.288, p = <.001, ηp^2 = .474). There was also a limb by time interaction (F (3.807,102.796) = 23.940, p = <.001, ηp^2 = .470)

Follow-up paired sample t-test revealed that slow limb time to P1 was greater than the fast limb at BL (p = .002, d = 2.04), however, slow limb time to P1 was less than fast limb when comparing limbs at EA (p = .016, d = .57). Slow limb time to P1 remained greater than the fast limb at LA (p = .006, d = .69).

The results of the follow-up paired sample t-test revealed that slow limb time to P1 decreased (p = <.001, d = 1.18) while the fast limb increased (p = <.001, d = 1.61) at the onset of the EA when compared to BL. Slow limb time to P1 exhibited no change at LA when compared to

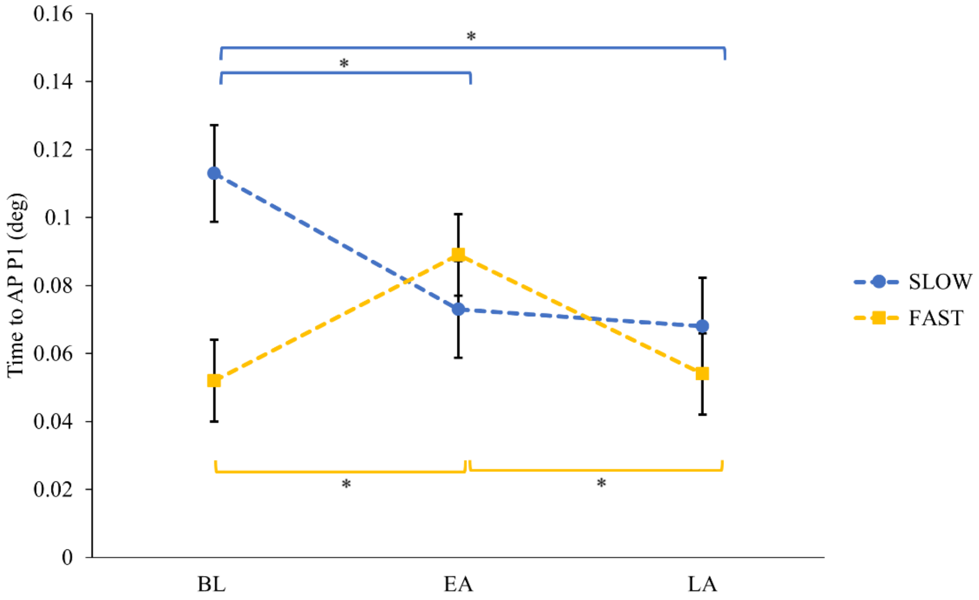
EA while the fast limb decreased (p = <.001, d = 1.46). Lastly, Slow limb time to P1 at LA was less than BL (p = <.001, d = 1.33), however the fast limb exhibited no difference.

Variability analysis revealed a main effect of limb (F (1,27) = 27.784, p = <.001, ηp^2 = .507) and main effect of time (F (3.186,86.029) = 44.418, p = <.001, ηp^2 = .622). Further a limb by time interaction was present (F (3.182,85.911) = 6.228, p = <.001, ηp^2 = .187).

Follow-up paired sample t-test revealed that slow limb time to P1 variability was greater than the fast limb at BL (p = <.001, d = 1.81), however, no difference was revealed when comparing limbs at EA and LA.

Follow-up pair t-tests found that slow limb revealed no difference in time to P1 at the onset of EA when compared to BL, while fast limb variability increased during EA (p = <.001, d = 2.25). Slow limb (p = <.001, d = 1.56) and fast limb (p = <.001, d = 2.02) time to P1 variability decreased when comparing LA to EA. However, slow limb time to P1 variability at LA was less than BL (p = <.001, d = 1.19) while fast limb time to P1 variability revealed no difference at LA when compared to BL.

Figure 17 – Slow & Fast time to AP P1 across BL, EA, and LA



* Denotes a significance at p = <0.017

Later Restabilisation Phase: (P2) Magnitude

The results showed that there was significant main effect of limb (F (1,27) = 398.180, p = <.001, ηp^2 = .936) and a main effect time (F (4.074,109.985) = 2.602, p = .039, ηp^2 = .088). There was also a limb by time interaction (F (5.690,153.625) = 36.415, p = <.001, ηp^2 = .574).

Follow-up paired sample t-test revealed that slow limb AP P2 was greater than the fast limb at BL (p = .002, d = .52) and greater than the fast limb at EA (p = <.001, d = 1.47). Slow limb AP P2 remained greater than the fast limb at LA (p = <.001, d = 2.96).

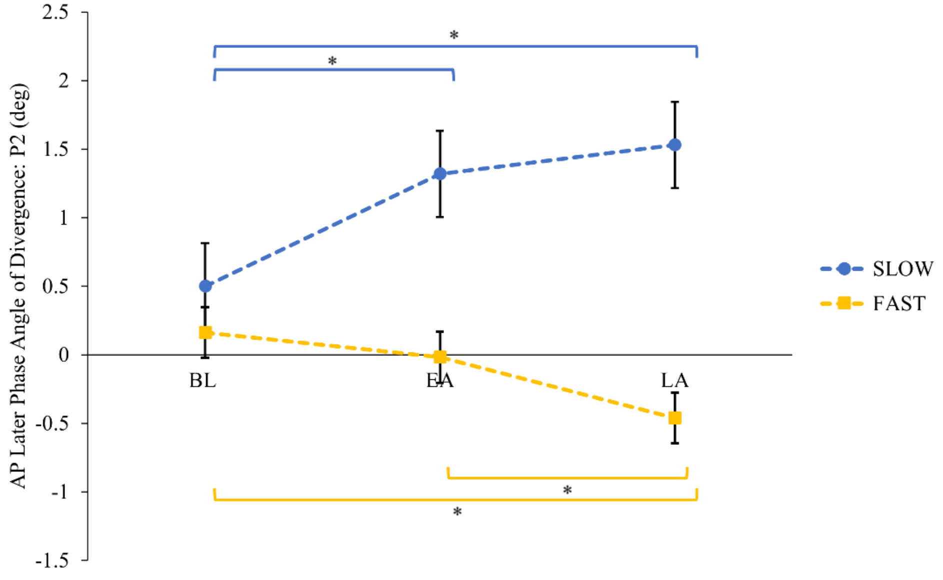
The results of the follow-up paired sample t-test revealed that slow limb P2 increased (p = <.001, d = 1.07) while fast limb P2 exhibited no difference at the onset of EA when compared to BL. Slow limb P2 revealed no difference at LA when compared to EA whereas fast limb P2 at LA decreased (p = <.001, d = .59) when compared to EA. Lastly, slow limb P2 at LA was greater than BL (p = <.001, d = 1.68) while fast limb P2 at LA was less than BL (p = <.001, d = .89).

Variability analysis revealed a significant main effect of limb (F (1,27) = 22.175, p = <.001, ηp^2 = .451) and a main effect time (F (4.034,108.926) = 34.194, p = <.001, ηp^2 = .559). There was also a limb by time interaction (F (5.439,146.848) = 4.877, p = <.001, ηp^2 = .153).

Follow-up paired sample t-test revealed that slow limb AP P2 variability was less than the fast limb at BL (p = .002, d = .93) but was greater than the fast limb at EA (p = <.001, d = 1.54). no limb difference was observed at LA.

The results of the follow-up paired sample t-test revealed that slow limb P2 variability increased (p = <.001, d = 7.02) while fast limb P2 variability also increased (p = <.001, d = 1.63) at the onset of EA when compared to BL. Slow limb P2 variability revealed a decrease (p = <.001, d = 3.44) at LA when compared to EA whereas fast limb P2 variability at LA decreased (p = <.001, d = 1.87) when compared to EA. Lastly, slow limb P2 variability at LA was greater than BL (p = <.001, d = .92) while fast limb P2 variability revealed no difference.

Figure 18 – Slow & Fast AP P2 across BL, EA, and LA



* Denotes a significance at p = <0.017

Time to P2

The results showed that there was a significant main effect of limb (F (1,27) = 108.238, p = <.001, ηp^2 = .800) and a main effect time (F (4.317,116.553) = 7.616, p = <.001, ηp^2 = .220). There was also a limb by time interaction (F (5.122,138.283) = 89.966, p = <.001, ηp^2 = .769).

Follow-up paired sample t-test revealed that slow limb time to P2 was greater than the fast limb at BL (p = .002, d = 2.72), however, slow limb was less than fast limb at EA (p = <.001, d = 1.95). Slow limb time to P2 remained less than the fast limb at LA (p = <.001, d = 2.09).

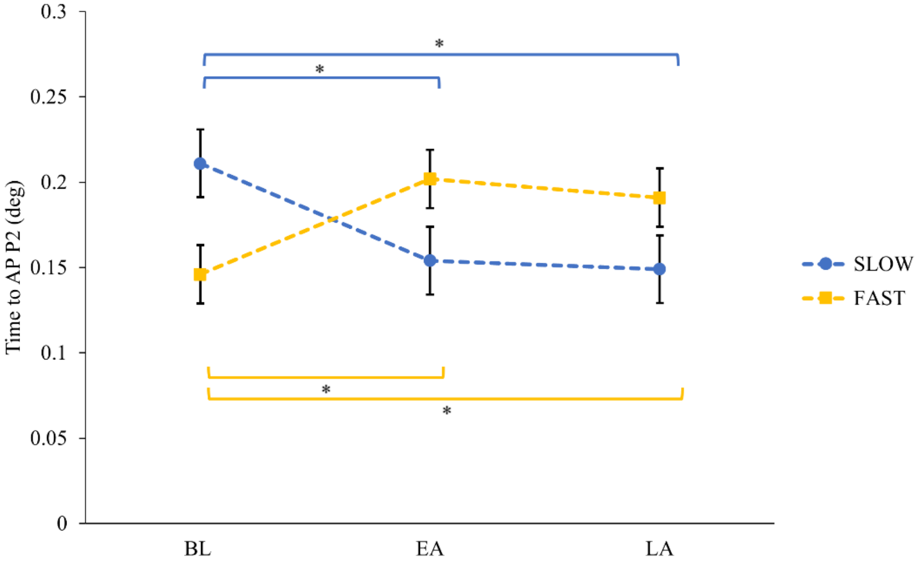
The results of the follow-up paired sample t-test revealed that slow limb time to P2 decreased (p = <.001, d = 2.44) while the fast limb time to P2 increased (p = <.001, d = 2.22) at the onset of the EA when compared to BL. Slow limb and fast limb time to P2 exhibited no difference at LA when compared to EA. However, slow limb time to P2 at LA was less than BL (p = <.001, d = 3.35) and fast limb P2 at LA was greater than BL (p = <.001, d = 1.79).

Variability analysis of time to P2 revealed a main effect of time (F (4.621,124.765) = 50.274, p = <.001, ηp^2 = .002). A limb by time interaction was found (F (5.066,136.785) = 10.712, p = <.001, ηp^2 = .284).

Follow-up paired sample t-test revealed that slow limb time to P2 variability was less than the fast limb at BL (p = .001, d = .88), however, slow limb variability was greater than fast limb at EA (p = .002, d = 1.21). No difference was revealed when comparing limbs at LA.

Additional analysis revealed that the slow limb increased variability (p = .002, d = 2.74) while the fast limb increased (p = .002, d = .72) when comparing EA to BL. Additionally, slow limb (p = .002, d = 2.63) and fast limb (p = .002, d = 1.37) decrease in time to P2 variability when comparing LA to EA. No significant differences were found when comparing the slow limb at LA to BL, however, fast limb variability was less than BL at LA (p = .009, d = .60).

Figure 19 – Slow & Fast time to AP P2 across BL, EA, and LA



* Denotes a significance at p = <0.017

***Table 12*– Comparison of anteroposterior kinetic variables between fast limb and slow limbs across all conditions**

	BL		**EA**		**LA**	
	SLOW	**FAST**	**SLOW**	**FAST**	**SLOW**	**FAST**
AP P1 (deg) ‡§	.718(.488)	2.255(.515)	2.645(.737)	.503(.589)	2.713(.583)	.742(.790)
Time to P1 (s) †‡§	.113(.041)	.052(.010)	.073(.025)	.089(.031)	.068(.025)	.054(.014)
AP P2 (deg) †‡§	.499(.510)	.163(.757)	1.32(.961)	-.018(.855)	1.53(.703)	-.461(.643)
Time to P2 (s) †§	.211(.019)	.146(.028)	.154(.027)	.202(.022)	.149(.018)	.191(.022)

*Data presented as mean (SD). † Denotes main effect time. ‡ Denotes main effect limb. § Denotes an interaction effect.

***Table 13* – Comparison of anteroposterior kinetic variabilities between fast limb and slow limbs across all conditions**

	BL		**EA**		**LA**	
	SLOW	**FAST**	**SLOW**	**FAST**	**SLOW**	**FAST**
AP P1 (deg) †	.222(.084)	.267(.084)	1.435(.632)	.642(.239)	.337(.121)	.351(.179)
Time to P1 (s) †‡§	.038(.024)	.007(.003)	.054(.033)	0.047(.025)	.015(.013)	.010(.007)
AP P2 (deg) †‡§	.216(.094)	.307(.101)	.923(.107)	.622(.254)	.362(.204)	.286(.010)
Time to P2 (s) †‡§	.013(.007)	.025(.018)	.069(.028)	.039(.021)	.013(.011)	.016(.011)

*Data presented as mean (SD). † Denotes main effect time. ‡ Denotes main effect limb. § Denotes an interaction effect.

Chapter 5: Discussion

Dynamic stability control during locomotion requires complex inter- and intralimb coordination to regulate the relationship between the position/velocity of the whole-body centre of mass within the base of support during both single- and double-support phases. Prior work investigating measures of dynamic stability during continuous gait perturbations (via split-belt protocol) have characterized temporospatial and kinematic control strategies and shown that such underlying components of dynamic stability can adapt to situational demands (Buurke et al., 2018; Darter et al., 2018; Liu & Finley, 2020; Park & Finley, 2017; Sawers et al., 2013). Additionally, work addressing the relationship between net eccentric ground reaction forces and COM kinematics has highlighted the importance of the magnitude and timing of applied forces as a means to offset angular accelerations during the double support phase of normal straight-line gait (Rawal et al., 2019). However, in the natural environment, such form of locomotion may occur relatively infrequently. Asymmetrical gait patterns, evoked through the use of split-belt treadmill, is one way to study the adaptability of the locomotor pattern. While considerable work has examined the temporospatial and kinematic adaptations that occur in the initial and later phases of the adaptation process, little research has explored force related measures specifically linked with stability and how these measures can inform us of the stability control strategies used when the limbs are required to perform different roles (i.e., in asymmetrical gait conditions). Therefore, the aim of the current study was to examine adaptive changes in the eccentricity of the ground reaction force during the double support phase of asymmetrical gait, with specific focus on mediolateral control. We specifically examined the initial proactive (P1) peak eccentricity along with the later reactive (P2) eccentricity as a means to better understand previously reported temporospatial and kinematic adaptations, which may be used to assist with the maintenance of mediolateral dynamic stability in asymmetrical gait. Furthermore, we also explored indices of mediolateral stability during asymmetrical walking using the margin of stability to understand how aforementioned kinematic, temporospatial and kinetic adaptations may influence stability. Secondary analyses concerning anteroposterior stability are also reported for completeness, but were not part of the initial research proposal.

Changes in Early Adaptation (EA) Phase Relative to Baseline (BL)

The first hypothesis predicted that during the initial delivery of an asynchronous gait pattern, participants would exhibit an increase in step width and step-to-step variability during initial delivery. During the EA phase (onset of asynchronous gait) fast limb step width did increase. Additionally, step-to-step variability increased during EA relative to BL.

The second hypothesis stated that during initial delivery ML MoS would increase while step-to-step variability in the ML MoS would increase. Our findings align with Buurke et al. (2018) as it was found that fast limb ML MoS increased during the EA. ML MoS variability revealed no significant change during EA relative to BL in the fast limb.

The third hypothesis stated that during the EA phase there would be no change in ML P1 magnitude or timing, and variability of the latter would increase. However, ML P2 magnitude and time to ML P2 would increase along with their respective variabilities. This was found to be partially true, fast limb P1 magnitude decreased while P2 magnitude increased in EA. Fast limb time to P1 and time to P2 remain unchanged. Both P1 and P2 magnitude variability increased in the fast limb, while both P1 and P2 timing variability exhibited no change in the fast limb during EA.

The aforementioned findings may suggest that during the EA phase increases in step width and fast limb ML MoS are a cautious control strategy to maintain stability when adapting to a novel gait pattern. Previous theory has suggested that although the nature of the task is in the AP direction, ML control strategies may be actively regulated (McGeer, 1990; Kuo & Donelan, 2010); therefore increases in step width, which may facilitate increases in later phase GRFnet, are done to actively maintain/increase ML stability.   Interestingly, while the P1 magnitude was always larger on the fast limb side (support that individuals scale initial GRFnet with gait velocity, a previously reported finding normal and fast overground gait (Rawal & Singer, 2021)), decreases in fast limb ML P1 magnitude in EA relative to baseline may suggest a need to maintain flexibility in control mechanisms rather than immediately matching proactive force output to the level of instability. As such, maintenance of stability on the fast limb side during a novel gait pattern may not require proactive changes in muscle activation to increase limb rigidity, a known contributor to initial peak GRFnet (P1) (King et al., 2014), but actively rely on later phase GRFnet eccentricities to maintain COM oscillations, seen by increases in P2 magnitude during EA relative to BL. The necessity or desire to maintain a larger margin of stability may have led to increased step width in the fast limb; although it remains difficult to decern the extent to which changes in the BOS strategy (i.e., step width) are driving the changes in force parameters. Given that P1 and P2 magnitudes changed differentially (and in opposite directions) during EA, it would appear that such force alterations may be, to some extent, independent of changes in the BOS.

Later Adaptation (LA) Phase Relative to Early Adaptation (EA) Phase

The fourth hypothesis considered that step width would remain unchanged relative to the EA phase, with step width variability decreasing relative to EA. This was found to be partially true, fast limb step width decreased when compared to EA values but remained greater than BL.

Additionally, decreases in fast limb step width variability were observed during LA relative to EA. Variability returned to BL levels.

The fifth hypothesis tested whether ML MoS would decrease relative to EA phase. Further, we believed participants would exhibit reduced ML MoS variability relative to EA with, but ML MoS variability would remain increased relative to BL. Similar to Buurke et al. (2018) fast limb ML MoS decreased during LA relative to EA. However, ML MoS was less than BL in the fast limb during LA. Additionally, ML MoS variability in the fast limb decreased during LA when compared to EA and was not different from BL values.

The final hypothesis stated that we would observe a greater P1 magnitude whereas time to P1 would remain unchanged and variability of both magnitude and timing would decrease, when compared to EA. Additionally, we believed the P2 magnitude and timing would decrease relative to the EA phase however it would remain higher than BL, with variability of timing and magnitude decreasing in the LA phase. It was found that fast limb P1 magnitude increased while P1 timing decreased when comparing LA to EA. Fast limb P1 magnitude was found to be greater than BL while P1 timing returned to BL levels. Fast limb P2 magnitude exhibited no change at LA when compared to EA and was still greater than BL. fast limb P2 timing exhibited no change when comparing EA to LA, and was still less than BL. Lastly, fast limb P2 magnitude variability decreased from EA to LA, to values similar to baseline.

The results suggest that some ML control mechanisms do exhibit adaptive qualities. Uniquely, fast limb ML MoS decreases to a value below BL, specifically during LA, likely a result of decreases in step width. Additionally, fast limb initial GRFnet peak (P1) occurs earlier in LA relative to EA and with a larger magnitude while no changes in later phase GRFnet are observed. This may suggest that when individuals are fully acclimated to a novel gait patten the greater reliance on reactive control measures diminish and proactive control strategies dominate in maintaining stability, a relationship previously observed by Rawal & Singer (2021) when there are constraints on foot placement in normal overground walking. This could also describe a scenario where individuals have solved control challenges of the novel gait pattern by matching applied forces to level of instability, allowing for smaller MOS without the potential for recurring instability. This may allude to the interplay between proactive and reactive control strategies in the presence of potential instability; nevertheless, changes in EA and LA during an asymmetrical gait pattern require each limb to complete separate tasks to maintain whole body stability. Therefore, going forward in this document an emphasis will be put on understanding differences between limbs at each discrete time point to understand the roles of each limb as they relate to dynamic stability.

Analysis of Mediolateral Stability

Intralimb Analysis: EA (Comparison Between BL & EA)

During the EA phase step width increases in both fast limb and slow limbs, may suggest a control strategy whereby individuals cope with a destabilizing task by increasing spatiotemporal control mechanisms to mitigate ML instability regardless of individual limb task differences. However, large increases in step width variability may allude to increased uncertainty during the initial delivery of an asynchronous gait and the necessity to control step-to-step fluctuations in COM position/velocity by modulating step width. Formerly is has been found that forward gait dynamics are passively regulated and that gait mechanics orthogonal to the direction of progression are actively regulated by the CNS (McGeer, 1990; Wurdeman et al., 2011). Therefore, initial changes in BoS parameters and increases in variability may suggest that individuals are actively controlling ML gait dynamics to preserve stability based on new information following each successive step. Further, it was found that ML MoS at the instant of slow limb heel strike exhibited no change at the onset of the EA when compared to BL while ML MoS at the instant of fast limb heel strike increased at the onset of EA compared to BL. Additionally, increases in variability were observed, however, likely a result of step width fluctuations. When coupled with the limb-specific changes in step width, results suggest a differential role of step width modifications in each limb. In the slow limb, increases in step width leads to no changes in ML MoS, which suggests stability is maintained by increasing step width, despite greater lateral COM displacements/velocities toward the slow limb side. In the fast limb, ML MoS increases may be due to a reduction in ML COM displacement/velocity, given that there was no differential step width change between limbs in EA. Consistent with this explanation, as fast limb belt velocity is larger (and consequently stance time would be shorter) there may be less time for the COM to move laterally, toward the fast limb, during double support. Therefore, an observed increase in fast limb ML MoS may be reflective of a strategy to increase stability by providing a greater buffer for potential instability that may arise on the fast limb side.

Force related stability measures would suggest that during the EA the slow limb and fast limb may be relying on reactive control strategies to maintain stability, but to different extents. The slow limb exhibits increases in both P1 and P2 magnitudes along with an earlier onset of P2, whereas the fast limb only exhibits an increase in P2 magnitude (which is greater than on the slow limb side). We may be able to infer that initial increases in the GRFnet eccentricity (P1) on the slow limb side are not perfectly scaled and therefore need to be supplemented by increases in reactive control measures (P2), even in the presence of step width (BoS) increases. Meanwhile on the fast limb side, increases in step width are accompanied by increases in the MOS, suggesting there may be less of a need for stability management early in the double support phase. Instead, residual instability can be managed later by the P2 eccentricity. While the P2 magnitude was greater in the fast limb than in the slow limb, the reduction in fast limb P1 during the EA phase (relative to BL) may suggest that individuals are relying less on preactivation of lower limb musculature, which may increase the necessity for reactive control strategies to maintain stability. Because BoS parameters increase on both sides while only ML MoS increases are noted in the fast limb, this may highlight an important relationship between proactive (P1) and reactive (P2) applied forces. These findings help build off previous work which suggest that P1 scales with gait velocity, that ML stability control may be mediated largely by foot placement and that the P2 eccentricity can help regulate angular momentum/stability control if instability persists during double support (Rawal & Singer, 2021). Inversely, if P1 modulation is sufficient, a dependence on P2 is less likely to ensure that COM motion is adequately maintained.

Perhaps, increases in P1 and P2 magnitude variability across both limbs suggest that individuals are experiencing some level of challenge scaling force output with instability on a step-to-step basis. Although, P1 and P2 magnitudes increased in the fast limb side relative to the slow limb, specific P2 increases in the fast limb during EA may be due to improper initial force modulation and therefore additional applied forces are required to ensure COM kinematics are adequately managed. However, more information may be needed to validate this alternate explanation; this in fact still alludes to the relationship between proactive and reactive control mechanisms as they tend to offset one another.

Intralimb Analysis: LA (Comparison Between EA & LA)

It was found that step width and variability in both limbs decreased in the LA. However, it was found that ML MoS decreased in both fast limb and slow limb when compared the EA phase; with a decrease in both fast limb and slow limb MOS variability. These findings are partially aligned with Buurke et al. (2018) whereby ML MOS in both limbs increase at EA and decrease during LA; this would further support the previous notion that during the onset of an asynchronous gait perturbation, ML gait dynamics reflect a conservative control strategy that promotes maintenance of stability. However, decreases in variability may suggest less uncertainty as a result of acclimation to the novel gait pattern, as individuals become accustomed to the COM dynamics and the associated force generation and modulation required to control the COM. However, over the adaptation period decreases in ML MoS coupled with decreases in step width may suggest that this sample of younger adults adapt to the new gait pattern and allow relatively greater amounts of COM movement within the narrower base of support limits. However, when interpreting force related measures, slow limb P1 magnitude decreased when comparing LA to EA whereas fast limb P1 increased; there were no changes in time to P1 nor changes in variability of timing, however variability of magnitude decreased in both limbs at LA. These findings may suggest that the fast limb adopts a more proactive level of control and therefore reductions in slow limb P1 over time may indicate that the fast limb is more engaged in proactive stability maintenance. Further, this may suggest that the greater fast limb initial force (P1) without decreases in later phase (P2) applied forces allows for the XCoM to move closer to BoS limits without surpassing lateral bounds – implying a less conservative control strategy. This adaptation may occur as individuals become acclimated to novel gait dynamics, that is, allowing the COM to exist closer to BOS limits, noted by the decrease in ML MOS in LA relative to EA. Further, P2 magnitude exhibited decreases in the slow limb without changes in the fast limb, whereas time to P2 remained unchanged in both limbs when comparing LA to EA. Magnitude variability and timing variability decreased in both limbs. These findings may suggest that although both step width and

ML MoS decrease, individual limb force modulation can adapt to task demands and become more effective, appropriately offsetting COM kinematics of a new gait pattern and rely less of reactive control strategies – seen by the decreases in P2 magnitude and variability.

Analysis of Anteroposterior Stability

Intralimb Analysis: EA (Comparison Between BL & EA)

When addressing spatiotemporal features in the AP direction it was found that step length exhibited adaptive changes. During the EA phase the slow limb exhibited an increase in step length while the fast limb decreased step length. This initial change in step length may be a result in a change in gait velocity; previous work has shown that when individuals are completing a split-belt treadmill protocol the body moves at a velocity that is similar to the average of the two independent belt speeds (Tesio et al., 2021). Therefore, the resultant gait speed of the split-belt protocol is faster than the slow speed BL and slower than the fast speed BL which results in the observed changes in step length during the EA phase. However, when assessing AP MoS in relation to step length we found that the slow limb and fast limbs exhibit a decreased AP MoS and increase in MoS variability. Distinctively, fast limb AP MoS becomes more negative during EA relative to BL. The aforementioned changes in AP MoS coupled with the initial changes in base of support suggest that there may in fact be some degree of instability on the slow limb side – that is, increases in base of support may be an indicator of a more safe support strategy however the decreases in AP MoS show that there is a greater amount of COM motion – bringing the XCoM closer to BoS limits. However, the fast limb undergoes a great deal of instability, a decrease in step length and a more negative trend in MOS during EA. This may suggest that reactive applied forces are imperfect and cannot offset increases COM accelerations.

Further, one could postulate that individuals at the onset of an asymmetrical gait pattern are falling onto the fast limb which may translate to the upper body being placed ahead of the lower limbs and therefore explain a negative AP MOS (XCoM lies outside BOS limits).

When interpreting force related measures, we found that the slow limb during the EA phase increased initial phase (P1) and later phase (P2) GRF_{net} along with decreased time to P1 and P2. Conversely, the fast limb exhibited decreased P1 and P2 magnitude along with increased time to P1 and time to P2. The following results would suggest that the slow limb responds to instability during EA by increasing step length and P1 magnitude to catch and manage excessive AP COM movements. The subsequent increase in P2 magnitude in the slow limb would suggest that the increases in P1 magnitude are insufficient and therefore require supplementation by the P2 response to offset COM motion is needed to ensure the system maintains stability. These findings coupled with decreases in time to minimum AP MoS following heel strike would further show that COM accelerations are increased, and the XCoM is approaching BoS limits at a much faster rate than what is seen during the BL trial and therefore require an increase in the initial response to instability. Further, considering the changes in fast limb dynamics, our results would suggest the fast limb is considerably less stable. The fast limb exhibits decrease in P1 and no significant change in P2 magnitude while fast limb AP MoS exhibited no change which is indicative of a control strategy that relies on reactive control measures to maintain stability when fast limb step length decreases – suggesting that COM movement may be less during fast limb support, due to less time in stance phase and therefore less time for the COM to travel. Additionally, increases in fast limb step time while the slow limb decreases (I.e., time to next successive step); we can postulate that the slow limb is responsible for the maintenance of whole-body stability as it spends more time in contact with the ground and therefore increases in force related measures of stability may be in part due to the lack of aid from the fast limb during EA.

Intralimb Analysis: LA (Comparison Between EA & LA)

During the LA phase, no adaptive changes in slow limb step length were observed, while fast limb step limb increased step length however remained less than BL measures. Although no changes in slow limb step length were present, slow limb AP MoS increased during LA; fast limb AP MoS also increased however was greater than BL measures. The results would suggest that individuals start to rely more on the fast limb, that is, the fast limb increases AP MoS likely as a result of step length changes. Previous work has reported an increase in limb symmetry when comparing EA to LA (Darter et al., 2018; Park & Finley, 2017). We may be able to infer that during LA, if limb symmetry increases, the individuals' interlimb stability control strategy become more evenly distributed. Further, fast limb step time also increases from EA to LA which may also suggest the fast limb is becoming more efficient at managing COM movements and therefore the slow limb does not need to be as involved during LA. However, during LA the slow limb also displays some adaptive mechanics; although no changes in step length are observed, there is an increase in AP MoS which may suggest that because the fast limb is increasing stability measures there may be less reliance on the slow limb to compensate and therefore more minimal changes are needed within the slow limb.

Additionally, during the LA the slow limb exhibited no change in P1 magnitude and time to P1, whereas the fast limb showed no change in P1 magnitude however a decrease in time to P1. Further, slow limb P2 magnitude and time to P2 exhibited no adaptive changes. while fast limb P2 magnitude decreased while an increased in time to P2 was exhibited. Lastly, variability decreased from EA to LA across all measures in both limbs. The following results would suggest that the slow limb displayed less uncertainty (decreased variability among all variables) over time which may be a result of acclimation to the novel gait pattern. The lack of slow limb adaptive changes in control strategies maybe a result of the fast limb using control strategies to counteract slow limb control challenges. Further our results highlight a previously reported relationship in gait mechanics and single step balance recovery (Rawal & Singer, 2021, Singer et al., 2019) in which the fast limb reverts to a proactive control strategy during LA, that is, no observed changes in P1 magnitude however a decrease in variability while P2 decreases. We can infer that during LA individuals use a control strategy that favors limb compliance at the instance of foot contact to produce adequate eccentric forces to produce a moment about the COM that offsets angular accelerations produced by the contralateral limb, rather than relying on later phase reactive forces (P2) to offset COM mechanics.

Limitations

The study collected data from a total of 28 healthy younger adults who ranged in stature. It is acknowledged that limitations arise as a function of differences in leg length and in preferred walking speed. Although fixed gait speeds were used to ensure changes in initial force magnitude (P1) were not a function of gait speed, in doing so this may skew BL stability parameters. If an individual has a slow or faster preferred gait speed in relation to either BL speeds, this may cause destabilizing effects and therefore stability mechanics may reflect such challenges. Further individuals who have a larger leg length may find the split perturbation to be less destabilizing than individuals with shorter leg lengths – differences in number of steps taken over the course of the trials. Additionally, based on Hof et al. (2007) a main contributor to AP

MoS is foot placement, therefore those with a larger length may take larger steps in which skew the results of base of support parameters. Although the analysis was completely within-subject, it is possible that such effects may increase the variability of the difference scores between conditions, complicating the detection of significant differences.

Another limitation of this study is anticipatory postural adjustments (APAs) which have been previously reported by Singer et al. (2014) in compensatory step protocols whereby if participants are aware of an imminent perturbation, changes in stepping parameters may be indicative of proactive strategies to preserve stability rather than relying on reactive measures. Further, participants may change the position of the upper body, that is, be urged to look down at the treadmill and feet as a strategy to overcome the continuous perturbation sooner. This may skew COM approximations as a large component of the COM calculations derived from the position of the trunk, arms, and head along with their respective kinematics.

Lastly, observed differences in EA (comparison between BL & EA) are comparing two different types of gaits: a steady state pattern and an asymmetrical pattern. Conclusions made from the EA should be limited as it may be unclear as to what changes are a function of decreased stability and what are requirements of the new gait pattern.

Conclusion & Future Direction

Gait is one of the most important and versatile types of locomotion used by humans. In the presence of instability, gait requires control of spatiotemporal characteristics to ensure the XCoM remains within support limits. The current work sought to understand the role of proactive (P1) and reactive control (P2) strategies that aid in the preservation of stability. It was found that initial stability was challenged such that changes in spatiotemporal parameters and changes in P1 & P2 timing /magnitude shift towards a more reactive strategy to control destabilizing COM movements. However, as individuals acclimate to a novel gait pattern, there is evidence to suggest a shift from reactive to proactive control strategies. Further our results suggest that in the ML direction individuals may in fact choose a control strategy in which both limbs encourage lateral safety while in the AP direction, individuals rely more on slow limb contributions to stability while the fast limb may only be used to help advance the system forward.

Our work has shown that there is a relationship between spatiotemporal characteristics and proactive/reactive related stability mechanisms in relation to COM motion. Future work should include clinical populations (i.e., older adult) to further understand the hierarchy that is stability control and acknowledge age-related changes in control strategies during continuous bouts of instability. The focus should remain on the relationship between spatiotemporal gait characteristics and force related measures as it pertains to their role in maintaining adequate MoS in the presence on continuous gait perturbations.

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Appendix A – Waterloo Footedness Questionnaire – Revised (WFQ-R)

Waterloo Footedness Questionnaire – Revised (Elias et al., 1998)

Instructions:

Please answer the following questions to the best of your ability. If you *always* use one foot to perform the described activity, circle **Ra** or **La** (for right always and left always). If you usually use one foot circle **Ru** or **Lu**, as appropriate. If you use both feet equally often, circle **Eq**.

1. Which foot would you use to kick a stationary ball at a target straight in front of you? Ra Ru Eq Lu La
2. If you had to stand on one-foot, which foot would it be? Ra Ru Eq Lu La
3. Which foot would you use to smooth sand at the beach? Ra Ru Eq Lu La
4. If you had to step up onto a chair, which foot would you place on the chair first? Ra Ru Eq Lu La
5. Which foot would you use to stomp on a fast-moving bug? Ra Ru Eq Lu La
6. If you were to balance on one foot on a railway track, which foot would you use? Ra Ru Eq Lu La
7. If you wanted to pick up a marble with your toes, which foot would you use? Ra Ru Eq Lu La
8. If you had to hop on one-foot, which foot would you use? Ra Ru Eq Lu La
9. Which foot would you use to help push a shovel into the ground? Ra Ru Eq Lu La
10. During relaxed standing, people initially put most of their weight on one foot, leaving the other leg slightly bent. Which foot do you put most of your weight on first? Ra Ru Eq Lu La
11. Is there any reason (i.e., injury) why you have changed your foot preference for any of the above activities? Ra Ru Eq Lu La
12. Have you ever been given special training or encouragement to use a particular foot for certain activities? YES or NO
13. If you have answered YES for either question 11 or 12, please explain:

Appendix B – Informed Consent

University of Manitoba Faculty of Kinesiology &
Recreation Management

RESEARCH PARTICIPATION INFORMATION AND CONSENT FORM

Research Project Title: UNDERSTANDING THE ROLE OF LOWER LIMB KINETIC ADAPTATIONS IN DYNAMIC STABILITY DURING NOVEL FORWARD WALKING

Investigator: Laine Dux (Graduate Student)
Study Coordinator: Jonathan Singer, Ph.D. (Associate Professor)

Sponsor: Natural Sciences and Engineering Research Council (NSERC) of Canada

This consent form, a copy of which will be left with you for your records and reference, is only part of the process of informed consent. It should give you the basic idea of what the research is about and what your participation will involve. If you would like more detail about something mentioned here, or information not included here, you should feel free to ask. Please take the time to read this carefully and to understand any accompanying information.

Purpose of this Study:
Falls during walking are a common and significant health concern for many older adults. These falls often lead to hip fracture, which reduces independent mobility and can increase the risk of mortality. Such reduction of individual independence can place a large burden on family members, caregivers and the Canadian healthcare system.

Despite considerable research, we lack specific understanding of why older adults fall. From a biomechanical perspective, the control of muscle force output from the lower limbs not only generates forward movement during walking, but also controls the movement of the whole-body centre of mass during a quiet stance and after a perturbation. This control is challenging because the combined force output from the lower limbs must first stop and then reverse the direction of centre of mass. Age-related changes in lower limb force output may change centre of mass movement, which can lead to instability.

The proposed work uses newly developed measures to better understand the biomechanical mechanisms underlying age in balance control during standing, walking, stepping and balance recovery following postural perturbation. Better understanding of the specific mechanisms that contribute to instability among older adults is important because it can improve the specificity of interventions to reduce age-related fall risk, such as targeted muscle strength, power training or perturbation-based training programs focusing on proactive or reactive control.

The objectives of the project are to: (a) determine force-related mechanisms underlying human stability control and instability among adults across a range of tasks; (b) link these challenges in control of balance to the level of individual joints of the lower limb; and (c) to discover the reactive mechanisms of continuous balance control measured by the technology within the laboratory.

Procedures Involved in this Study:
Being screened for falls during normal activities, age, head injuries, etc. to see if participants are eligible to participate in the study.

This study consists of:

Experimental Set-up:

Anthropometric measurements (height and body weight) will be recorded at this time for use in the biomechanical model (the biomechanical model is a mathematical model of your skeleton that we use to determine the position of your whole-body centre of mass and the force transmitted across your joints).

Lower body reflective markers will be placed on specific boney landmarks around the hip (front, back, and side), additionally markers will be placed on the upper thigh, knee, and ankle; while the remaining markers will be placed on the heel, forefoot, and toes. Upper body reflective markers will be placed around the chest, shoulders, elbows, wrists, and hands; with additionally markers places on the ear lobes. Rigid clusters containing four markers will be placed on the head (using a headband), low back, upper back and on both upper arms, lower arms, lower legs, and thighs. The Reflective markers will be placed on top of your clothing or directly on the skin using double-sided tape.

You will be asked to stand upright and as motionless as possible, with your arms at your sides, for approximately 5 seconds. We will record the position of all the reflective markers on your body. This information will be used to build the biomechanical model (a mathematical representation) of all your body segments (e.g. arms, legs, trunk, head) in space.

Walking:
This condition will be done split-belt treadmill with a researcher close by. You will take part in a test for limb preference via the Waterloo Footedness Questionnaire. Prior to data collection you will be required to complete a familiarization period on the split belt treadmill. The familiarization period will have you walking at two different gait speeds; the first will be the slow period in which both belts will operate at 0.5m/s, you will walk at this speed for 2 minutes. You will then walk at a fast

speed (i.e. 1 m/s) for an additional 2 minutes. Lastly, baseline measures will be collected with the belts in tied mode at 1.5m/s velocity. Baseline measures will be collected for 10 strides.

Upon completion of the familiarization phase a continuous treadmill perturbation will be delivered. The right belt will be set to 1.5 m/s while the left belt will be set to 0.75 m/s. You will walk continuously with the treadmill belts uncoupled for 15min. During this block specified data (kinematic and kinetic) will be repeatedly collected at 1min intervals starting at the onset on the belt perturbation. Outcome variables will be collected as an average over 10 strides at each respective time interval.

After the completion of the testing period you will go through a 5 min washout period in which the belt will be re-coupled (set to 0.75 m/s) to allow you time to safely return to a normal gait patten. You will have the option to extend the washout period as needed to ensure you are comfortable with coupled walking upon leaving the laboratory setting.

You will be asked to wear your own shorts, t-shirt and running shoes while participating in the study. You can choose to take part in any or all conditions. The experimental set-up and retroreflective marker placement takes approximately 45 minutes; walking trial requires 30 minutes. Total Study time is ~ 1.5hr. You have the choice to participate in any or all conditions.

Eligibility Requirements/ Screening Procedures:
To be eligible for the study, you must not be allergic to the prophylactic taping adhesive. You should have no diagnosed lower body anatomical, neurological impairment, or injury affecting gait during testing within the last 6 months. You should have not had any lower body surgery within the one year. You should be able to stand and walk without aid for up to 1 hour. You should have no previous history of falls during normal activities of daily living. You should not be taking any psychoactive medication for at least a week before recruitment. You should be able to understand and follow instructions to rule out cognitive impairment. Please understand the years of age of interest is 18-30 years of age.

Recording Devices:
During all trials, a motion analysis system will record the position of each reflective spherical marker you have placed on your body. The cameras that record the position of these reflective markers only respond to infrared light and are not capable of recording images of anything other than the reflective markers (i.e. it is not possible to see images of your person, as you would see with a typical video camera). The information we obtain from the position of these reflective markers is fed into the biomechanical model and used to compute the position of your whole-body centre of mass and the forces that are transmitted across your joints.

We will also record the forces that you exert on the ground, using force platforms imbedded in the split-belt treadmill. A force platform is similar to a typical bathroom scale, except that a force platform also responds to forces applied in the front-to-back and side-to-side directions, in addition to forces in a downward direction.

Benefits of Participation:
Apart from the opportunity to learn about how humans control their balance during standing, walking, and after being perturbed, there are no direct benefits to you from participating.

Risks to Participation and Associated Safeguards
If you are unable to react appropriately to the external perturbations or adverse conditions presented during the delivery of an asymmetrical gait pattern, you may experience unstable gait. Additionally, this study requires the use of fixed and uncoupled belt speeds - these speeds may be faster than the participants everyday walking speed and as a result cause some uncertainty and fear of loosing balance during walking. In some individuals, the adhesive tape used to affix the reflective markers to the skin has caused some redness and discomfort.

As you are walking on a treadmill with fixed speeds which may be faster than your average walking speed and are needed to walk with the treadmill belts operating at two different speeds there is a risk of loosing balance and falling. Risks associated with the treadmill and the delivery of a asymmetrical gait patterned will be minimized by using a standard safety harness which the you will be required to wear. The safety harness is connected to a rigid steel frame overhead. The length of the cable connecting the harness to the rigid frame will be set to such a length that in the event you may fall, you will not come in contact will the treadmill but be suspended in the air. Additionally, the testing treadmill has railings along the sides and in front of you in the event you require something to grasp onto. You will also have a emergency tether attached to your waistband - if you move too far back on the treadmill the tether will release bringing the treadmill to a stop.

In the event that you need to stop, an emergency stop button will be mounted on the railing of the treadmill. Additionally, the investigator will have an emergency stop button to ensure your safety. A research assistant or I will be constantly observing you.

In some individuals, the adhesive tape used to affix the reflective markers to the skin has caused some redness and discomfort. If you begin to experience skin irritation as a result of the adhesive tape, you will be asked if you wish to continue with the study. If you do wish to continue with the study, you will be asked to make one of the investigators in the room aware if you begin experiencing additional redness or discomfort at the sites of the reflective markers. In such cases, testing will stop immediately. If, during testing, you notice that you are experiencing redness or discomfort at the sites of the reflective markers, please make of the investigators in the room aware of this and testing will stop immediately.

Anonymity and Confidentiality of Data:
As the motion analysis cameras only record the position of the reflective markers located on your body, this data is completely anonymous. There is also no way to identify you from the forces you apply to the force platforms. You will be identified only by a participant identification code, which contains no personally identifiable information. These codes contain only a number/letter and cannot be linked back to any specific person.

This consent form and the bottom portion of the participant feedback form, which will contain your name and signature, will be kept in a locked filing cabinet in the principal investigator’s office for three years after the completion of the study (the principal investigator's office is located behind two locked doors). Only the principal investigator and advisor will have access to the consent and feedback forms. After this time, the consent and feedback forms will be destroyed via a file shredder (07/2023).

The principal investigator, research assistant, and advisor will have access to biomechanical data (motion analysis, force platform and treadmill data) collected during this study. These data do not contain pictures of individuals or any other information that could be used to identify a participant information. The biomechanical data, which are anonymous, will be retained indefinitely on a password-protected computer in the principal investigator's lab (biomechanics lab).

Remuneration:

Upon arriving to the university, we will have provided a parking pass to cover the cost of your parking (Lot X). If you took public transportation to the university, you will be reimbursed for the cost of public transportation to the university and the return trip home for each time you come back to participate in the study.

Changing Your Mind about Participation

You may withdraw from this study at any time without any negative consequences. You may withdraw before the study; you may do so by email/ phone/ in-person or during the study. To do so, indicate this to the researcher or the research assistant by saying or emailing, 'I no longer wish to participate in this study', or a similar statement. If this statement was provided before or during the study, we will delete/shred any hard copy information and delete the motion capture and force platform data.

If you no longer want your data to be used, please note down the date and time of your participation, as this is the only way to locate your data and delete it. Please note that there will be an inability to withdraw any data collection by January of 2023. At this point in time, all data collected will be used for discussion and be analyzed to view level of significance. The data will then be published and at that time be public data. However, no personal information will be included to identify the participant.

Participant Feedback

After your participation in the study, you will have the opportunity to discuss the research with the principal investigator, should you have any questions. As the raw data collected take some time to process in order to obtain any interpretable results, you will have the option of indicating that you would like a summary of the research results following the completion of the study. Results should be available during the January of 2023. On the participant feedback form, you will be able to provide an email address to which we will send the results.

Dissemination of Results:
Results of this study will be presented at academic conferences (such as the Congress of the Canadian Society for Biomechanics or the International Society for Posture and Gait Research). Data will also be published in manuscript format (such as in the Journal of Biomechanics, the journal Gait and Posture, and MSpace). Data will be presented as group average values – there will be no information presented that could identify you as a participant in this study. All data remains confidential.

Your signature on this form indicates that you have understood to your satisfaction the information regarding participation in the research project and agree to participate as a subject. In no way does this waive your legal rights nor release the researchers, sponsors, or involved institutions from their legal and professional responsibilities. You are free to withdraw from the study at any time, and /or refrain from answering any questions you prefer to omit, without prejudice or consequence. Your continued participation should be as informed as your initial consent, so you should feel free to ask for clarification or new information throughout your participation.

The University of Manitoba may look at your research records to see that the research is being done in a safe and proper way.

This research has been approved by the Research Ethics at the University of Manitoba, Ft. Garry Campus. If you have any concerns or complaints about this project you may contact any of the abovenamed persons or the Human Ethics Coordinator at 204-474-7122 or humanethics@umanitoba.ca. A copy of this consent form has been given to you to keep for your records and reference.

Participant's Signature ________________________ Date ____________

Researcher and/or Delegate's Signature ___________________ Date ___________

Appendix C – Inclusion/Exclusion Criteria

**These are questions that will be asked during initial screening and the list is to ensure the PI asks all questions to all participants in a consistent manner.

Inclusion/Exclusion Criteria Checklist

1. How old are you? ____________ (18-30, 40-55 or +65 years?)

2. Do you require an aid (e.g., walker, cane) to stand or walk?

3. Have you had a fall during the last year while performing activities

4. of daily living (walking, household chores, etc.) that has affected your mobility?

5. Have you had a lower body injury within the last 6 months?

6. Have you had any lower body surgeries in the last year?

7. Do you have a diagnosed anatomical, neurological, cognitive, or visual condition or injury that affects your ability to walk on a treadmill?

8. Are you currently taking any substances or medications with known on your balance control?

9. Do you have a known allergy to latex or have experiences skin irritation when using bandages?

10. Did you answer 'Yes' to any question on the COVID-19 Screening tool?

A 'Yes' answer to any question is a reason for EXCLUSION for this study.

Appendix D – Demographics Questionnaire

Participant Code: _______________

Demographics Questionnaire

Age: _______________

Height: _____________

Weight: _____________

Sex assigned at birth: O Male

O Female

O Prefer not to say

Gender: O Man

O Woman

O Transgender

O Nonbinary

O Other (please specify): __________________

O Prefer not to say