Examining the biomechanical mechanisms of restabilisation-phase stability control of compensatory stepping with increasing perturbation magnitudes and with failed balance recovery responses

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Abstract

PROBLEM: Falls are an ever growing problem in Canada that can lead to physical injury. Falls occur when the Center of Mass (CoM) moves outside the borders of the Base of Support (BoS), putting the body in an unbalanced state. Following a balance perturbation, a compensatory step acts reactively to prevent falls via a forward step that increases the BoS. After foot-contact, the restabilisation phase creates stabilizing moments that act to slow and maintain the CoM within the BoS. However, no studies have looked at the how the restabilisation phase changes with increasing perturbation magnitudes, and when a compensatory step is unable to recover balance. **PURPOSE:** To assess the biomechanics of the compensatory stepping response and its restabilisation phase with increasing perturbation magnitudes, as well as when the compensatory step is unable to recover balance with a single step. **METHOD:** A maximum lean-and-release protocol was used to compare (1) participants successful balance responses at their initial and maximal lean angles and (2) successful and failed balance response tasks that occur at the same lean angle. **RESULTS:** Successful balance responses following larger perturbation magnitudes had a reduced step time, longer step length, and larger stabilizing moments during the restabilisation phase, but had a larger destabilizing moment at foot-contact and greater instability throughout the restabilisation phase, as well as a longer time to restabilisation. The failed balance response tasks relative to the successful responses had shorter step lengths, larger destabilizing moments and greater instability at foot-contact, and were unable to achieve restabilisation. **CONCLUSION:** The key to performing a successful balance recovery response appears to be producing a large and fast step that reduces the CoM velocity at foot-contact, increases stability at foot-contact, and allows for generating larger stabilizing moments in the restabilisation phase.

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Introduction

There exists a large array of research concerning falls whether it regards the incidence rates, risks, physical and psychological harm, or the differences in the aforementioned topics across the lifespan. Despite all this research, it is still not specifically understood why these falls occur. This is especially disconcerting when looking at the impact of falls on society. While falls represent the leading cause of injury for all ages, older adults are found to have the greatest incidence rates and negative outcomes. It has been documented that about one-third of community-dwelling older adults over the age of 65 years and approximately half of institutionalized adults over the age of 80 will fall each year (Tinetti, Speechley, & Ginter, 1988). Falls account for 85% of all injury-related hospitalizations among Canadian seniors as well results in the longest length of stay compared to other causes of injury (Stinchcombe, Kuran, & Powell, 2014). Furthermore, injuries sustained from a fall act to further increase one's risk with nearly half of older adults who have fallen experiencing a repeat fall within the next year (Gill, Murphy, Gahbauer, & Allore, 2013; Tinetti et al., 1988). While many falls do not result in any injury, about 31% of falls will require medical attention with 10-15% of falls resulting in fractures and 5% of falls resulting in more serious soft tissue damage or head trauma (Stevens, Mack, Paulozzi, & Ballesteros, 2008; Tinetti, Doucette, Claus, & Marottoli, 1995). Falls have also been associated with mortality with this relationship strengthening with age (Bhattacharya, Maung, Schuster, & Davis, 2016). Even if a fall is not a direct cause for death, the resulting injuries may precede such an outcome. Although only a result of 1% of falls, hip fractures are a prime example of such severe consequences as disability or death (Hayes et al., 1996). The consequences of falls extend even beyond physical harm resulting in decreased independence, self-efficacy, nutrition, and an increased fear of falling (Stel, Smit, Pluijm, & Lips, 2004; Tinetti & Williams, 1998). This fear of falling is problematic as it derives from a history of falling yet also increases the risk for future falls, creating a dangerous cycle (Young & Williams, 2015). Along with fear, there are plenty more factors that increase this risk. Specifically, falls are especially prevalent among individuals with a medical history of a stroke, poor nutrition, and obesity as well as the age-related consequences of decreased muscle mass, lengthened reaction time, arthritis, and bone mineral density loss (Chang & Do, 2015). Though these factors increase the risk of falls, they are not the reason why people fall in the first place. Therefore, the purpose of the following work is to better understand the underlying biomechanical mechanisms that contribute to falls.

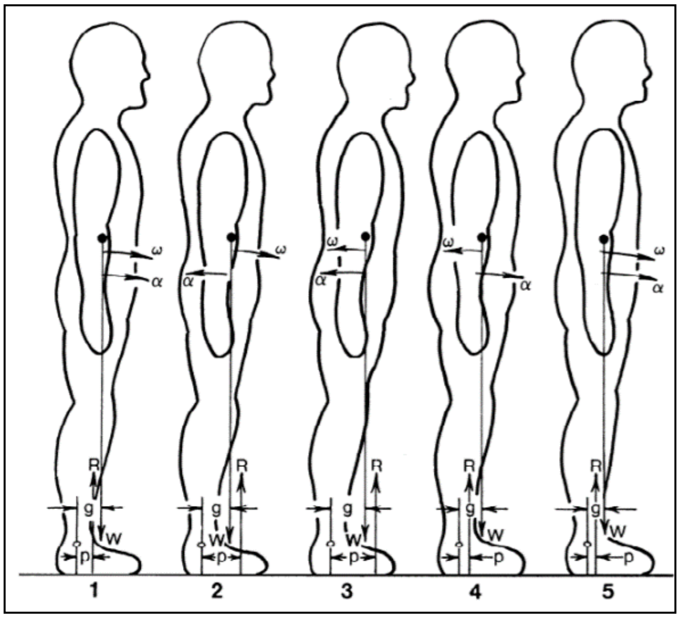
Mechanisms of Balance Recovery Responses

To understand the mechanics of a fall, it is important to know the relationships between various balance-related variables. While the center of mass (CoM), base of support (BoS), and center of pressure (CoP) are independent quantities, the regulation of one variable can significantly influence the state of another. The balance control system, for example, regulates the relationship between the CoM and the BoS (Maki & McIlroy, 1996). Balance, or equilibrium, can be defined as the state of an object when the resultant forces and moments acting on it are zero. In relation to humans, balance is dependent on the location of the vertical projection of the CoM in relation to the area of the BoS (Winter, 1995). When considering the instantaneous relationship between the CoM and BoS, if the CoM falls within the BoS, balance is achieved, however, it is when the CoM extends beyond the boundaries of the BoS that the body becomes unbalanced (King, Judge, & Wolfson, 1994; Lockhart, Smith, & Woldstad, 2005). It is in this state of imbalance that if the body is unable to restabilise, a fall will occur. Stability, in relation to human balance control, refers to the ability to maintain, achieve, or restore a state of balance, and seeks to do this by regulating CoM kinematics either through appropriate ground reaction force generation – to alter CoM kinematics – or by changing the BoS – to afford a greater potential to recapture the CoM (Horak, 1987; Singer, Prentice, & McIlroy, 2016). Disruption of the CoM-BoS relationship can occur due to an external perturbation - a disrupting force from outside the body (such as being pushed or tripped) - or an internal perturbation - a disrupting force from within the body (such as raising one's arms or initiating gait) (McIlroy & Maki, 1999). Regardless of the perturbation method, when the body is in a state of imbalance some form of compensatory forces and movements must be made to restabilize the body.

Balance recovery tasks from a standing posture include a large range of movements from simply plantar- and dorsiflexing at the ankle to taking multiple steps. These can be broken into two classes of strategies – fixed-support and change-in-support strategies (Maki & McIlroy, 1997). The differentiation between these strategies being whether limb movements occur to alter the BoS. Fixed-support strategies keep a constant BoS and alter the CoP location in an attempt to keep the CoM within the BoS. Change-in-support strategies alter both the BoS and the CoP location in an attempt to redirect and maintain the moving CoM within the borders of the new BoS. The CoP-CoM relationship can be explained by the Inverted Pendulum Model (Winter, 1995; Winter, Patla, Prince, Ishac, & Gielo-Perczak, 1998; Winter, Prince, Frank, Powell, & Zabjek, 1996). An inverted pendulum occurs when a pendulum has its CoM superior to its pivot point. During quiet standing in humans (Figure 1), the ankle joint acts as the pivot point with two external forces acting about the ankle: the bodyweight (W, which acts at the CoM) and the vertical ground reaction force (R, which acts at the CoP). Because, during quiet standing, the

Figure 1.1

The Inverted Pendulum During Quiet Standing



Note. From Winter (1995).

magnitude of the bodyweight is equal and opposite to the vertical ground reaction force, if the CoM and CoP are at an equal horizontal distance of g and p, respectively, from the ankle, the two external moments acting about the ankle will be equal, keeping the body in static equilibrium. Alternatively, any differences in the horizontal locations of the CoM and CoP from the ankle will result in a destabilizing (or restabilizing) moment that accelerates the CoM in the opposite direction of the CoP (Winter, 1995; Winter et al., 1998, 1996). Because of this, balance response tasks aim to displace the CoP such that it can reverse the polarity of the CoM acceleration, and consequently the CoM velocity, keeping the CoM within the BoS.

Two common fixed-support strategies include the "ankle strategy", effective for small perturbations, and the "hip strategy" which can be used for slightly larger perturbations. For anteroposterior perturbations, the ankle strategy involves contracting the plantarflexor or dorsiflexor muscles to shift the CoP anteriorly or posteriorly, respectively, beyond the CoM to reverse the polarity of its acceleration (Winter et al., 1996); this is the strategy used in Figure 1. The hip strategy differs in that the hip flexors or extensors activate in response to anterior or posterior perturbations, respectively, to produce shear forces that oppose the direction of the perturbation and move the CoM posteriorly or anteriorly, respectively (Winter et al., 1996).

Mediolateral perturbations are unique in the sense that both the ankles and hips work together, but to varying degrees. This is due to the loading/unloading mechanism in mediolateral balance control. To shift the CoP in the ML direction, the hip adductors and opposite-limb abductors contract simultaneously to increase the load over one limb while simultaneously decreasing the load by an equal amount over the opposite-side limb; concurrently, the respective evertors of one ankle and the invertors of the opposite-limb ankle contract simultaneously to aid the hips in shifting the CoP mediolaterally. This results in the CoP quickly shifting closer to the loaded limb (Horak & Nashner, 1986; Winter et al., 1996). Because fixed-support strategies require the CoM to remain within the BoS, the ankle and hip strategies are only effective at maintaining balance for small perturbations; even then, change-in-support strategies often occur from similar perturbation magnitudes, likely to safeguard stability (Maki & McIlroy, 1997). For change-in-support strategies, the BoS is altered either by taking a step or grasping an object. Change-in-support strategies are more advantageous than fixed-support strategies for two main reasons (Maki & McIlroy, 1997). First, they create a larger base of support; this allows for a greater displacement of the CoM without losing stability. Second, there is a larger moment arm between the foot-/hand-contact force and the CoM; these stabilizing moments act to decelerate the CoM. Because of these advantages, change-in-support strategies provide the only responses able to recover from large perturbations – which are typically performed as reaching to grasp for the upper extremity or, more commonly, compensatory stepping for the lower extremity (Maki & McIlroy, 1997). Reaching to grasp, while an excellent method in preventing a fall, can only occur when there is something to grab onto whereas compensatory stepping can be performed on any relatively level ground. Additionally, change-in-support reactions have the advantage of being initiated and executed very rapidly after the onset of instability, especially when comparing compensatory stepping to volitional stepping (Burleigh, Horak, & Malouin, 1994; McIlroy & Maki, 1993; McIlroy & Maki, 1996). For these reasons, compensatory stepping is generally efficacious in preventing a fall.

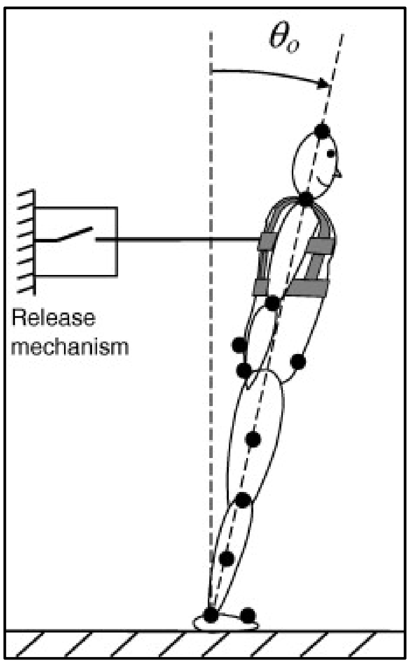
A compensatory step can be broken down into three phases: (1) step initiation/preparation phase, (2) swing phase, and (3) contact/restabilisation phase (Do, Breniere, & Brenguier, 1982; Hsiao & Robinovitch, 1999; King, Luchies, Stylianou, Schiffman, & Thelen, 2005). The step initiation/preparation phase, which occurs from the perturbation onset until toe-off, involves reacting to the onset of instability and preparing for the swing phase. It is in the events leading up to toe-off of the preparation phase that the mechanics of the swing phase are determined, which occurs between toe-off and foot-contact. These include the step execution time (time from toe-off to foot-contact) and step length, both relying on the propulsive forces produced by the swing/stepping leg (Do et al., 1982). In a well-practiced compensatory step, the presence of anticipatory postural adjustments (APA) may be utilized in the step initiation phase which can also influence swing phase mechanics (Mcllroy & Maki, 1995). The APA (a rapid movement of the CoP toward the swing-limb) results in a shifting of the CoM towards the swing-phase-BoS (stance leg) prior to toe-off, which influences the lateral displacement of the CoM during the swing phase. The closer the CoM is to the stance leg at toe-off, the less lateral displacement the CoM will experience during the swing phase, as a result of the external gravitational moment about the stance-limb BoS (i.e., ankle joint centre). However, compensatory stepping responses are often too quick for an APA to occur or to have a substantial effect on the mediolateral displacement of the CoM (McIlroy & Maki, 1999), resulting in increased lateral (toward the swing limb) displacement of the CoM during the swing phase. The swing phase ends at foot-contact in which the restabilisation phase begins. This phase is important for the maintenance of dynamic stability and is achieved by the CoP being anterior and lateral to the CoM, causing a deceleration and eventual stop of the CoM (Singer et al., 2016). Upon the CoM stopping within the BoS, the body becomes restabilized and the possibility of a fall diminished. In cases where the velocity of the CoM is too high and unable to decelerate in time, or there is a high level of instability during the restabilisation phase, another step may be necessary to prevent a fall. To undertake research in preventing falls, it is often necessary to induce a state of instability to observe and analyze how people perform such compensatory movements.

Methodologies to Study Dynamic Balance Control

Many different experimental paradigms have been used to induce a perturbation or loss of balance which requires a balance recovery response. Such paradigms may include participants walking on a slippery surface (Bhatt & Pai, 2009; Troy & Grabiner, 2006) or into unexpected obstacles (Hinkel-Lipsker, Stoehr, Lachica, & Rogers, 2021; Pijnappels, Bobbert, & Van Dieën, 2004). Many paradigms have also included balance recovery tasks starting from an upright stance. These include being tugged at the waist by a cable (Luchies, Alexander, Schultz, & Ashton-Miller, 1994; M. W. Rogers, Hedman, Johnson, Cain, & Hanke, 2001), a sudden translation of the floor (McIlroy & Maki, 1996; Pavol, Runtz, Edwards, & Pai, 2002), or being released from a support cable while leaning forward (Do et al., 1982; Singer et al., 2016). While examining falls that occur from walking will allow for high external validity of common causes that may induce a state of imbalance, it is challenging to determine the specific mechanisms that result in instability when recovering balance as one step can cause instability in future steps, making it hard to determine a cause and effect relationship. Alternatively, balance recovery tasks from an upright stance allow for a more in depth examination of those mechanisms of compensatory stepping, as well helps to inform researchers of the ease or difficulty of different populations' ability to recover from perturbations. For these reasons, this research paper will focus on the paradigms that start from an upright stance.

Figure 1.2

The Lean-and-Release Protocol Prior to Cable Release



Note. From Hsiao-Wecksler (2008).

The lean-and-release protocol (Figure 2) simulates an unbalanced body at the beginning of a trip (Hsiao-Wecksler, 2008). This is achieved by a participant being held in a static forward or backward lean via a horizontal cable attached to the participant by a chest harness or waist belt. When at the desired lean angle, the cable – at a pseudorandom time delay – is released, causing a loss of stability and/or balance (Do et al., 1982). As the lean angle increases, so too does the perturbation magnitude acting on the participant which requires a more substantial balance recovery response (Hsiao-Wecksler & Robinovitch, 2007). At low magnitude lean angles, the ankle or hip strategies may be enough to regain stability. With greater lean angles/higher perturbation magnitudes, and the subsequent loss of balance, a compensatory step will be required to recover balance and achieve stability (Mackey & Robinovitch, 2006). A single compensatory step is often enough for sub-maximal lean angles though if the perturbation is large enough a multi-step strategy may be necessary to recover balance. This paradigm is especially useful when the starting conditions of each trial must remain the same (i.e., beginning each trial from the same position with little-to-no movement prior to stepping) as this allows for a more in depth examination and comparison of the mechanics within and between participants.

The cable pull paradigm differs in that instead of a cable releasing a participant, it is pulling on them. Participants begin in an upright, static position with multiple cables attached to their waist via a rigid ring or padded belt in the anterior, posterior, and both lateral aspects (Rogers et al., 2001; Schulz, Ashton-Miller, & Alexander, 2005). Each cable is attached to its own designated weight unseen from the participant with magnets supporting said weights. Researchers are able to disengage a magnet, leading to the weight dropping and the cable pulling the participant in one of the four different directions (Rogers et al., 2001; Rogers, Hedman, Johnson, Martinez, & Mille, 2003). Alternatively, the cables may be pulled by a motor that allows for more control in perturbation magnitude between and within subjects (Mille et al., 2013). With the cable being pulled, this creates a perturbation causing a rapid acceleration of the CoM that necessitates a compensatory response. The perturbation magnitude can be scaled by increasing the weight pulling the cables or by increasing the speed of the motor (Rogers et al., 2001; Rogers et al., 2003). This paradigm is beneficial as it allows for perturbing participants in multiple directions, unpredictability in perturbation direction and timing, is consistent in the force applied between- and within-subjects, and allows for easy scaling in perturbation magnitude.

Surface translations differ from the other paradigms as they produce a perturbation at the feet. This is achieved by a rapidly moving surface platform that a participant is standing on, which is able to move in multiple directions (anterior, posterior, and lateral) (Mansfield & Maki, 2009). As the feet move with the platform, the inertial properties of the CoM causes it to lag behind, putting the body off balance and thus necessitating a balance response. The magnitude of the perturbations can be gradually increased via an increased distance, velocity, and/or acceleration of the moving platform (Tokuno, Cresswell, Thorstensson, & Carpenter, 2010). A major benefit to this methodology is its ability to detect smaller differences between age populations, however it has been shown that the different perturbation methods produce similar results (Lakhani, Mansfield, Inness, & McIlroy, 2011; Mansfield & Maki, 2009).

With many different methodologies for inducing a perturbation, it is important for researchers to determine which one best helps them to answer their research question. Each of these methodologies share similar benefits. One shared benefit is their high internal validity as each method allows for a controlled balance response task that can be simulated similarly across trials and researchers (Mansfield & Maki, 2009). The cable pull and surface translation are both able to produce perturbations in four directions while the tether-release protocol can be used to simulate both anterior and posterior falls. The main downside to these paradigms is that they are lower in external validity compared to methods that have participants walking prior to being perturbed, as most falls occur during locomotion (Tinetti et al., 1988). However, to the benefit of surface translations and cable pulls, participants are able to move around, even if only slightly, prior to being perturbed; this allows for more freedom in experimental designs when using these methodologies. Though, depending on the research question, it can often be valuable to induce a loss of balance from a static position. This is often the case when the purpose is not to examine the cause of the initial instability but rather the performance of the balance response task itself. By limiting the movement prior to a perturbation, trials are able to be more consistent and allow for a more in-depth look at the mechanics of the balance response task without worry of the influence that prior movement may have on the outcome variables. This is especially beneficial in the tether-release protocol as participants begin each trial from the same position with little-to-no wiggle room. This allows for analysis of the compensatory step with as few confounding variables that may affect its mechanics. The lean-and-release protocol is not without its downsides though as the direction and magnitude of perturbations are highly predictable and participants have been shown to gain practice-related improvements in the kinetic and kinematic measures of their compensatory step restabilisation (Singer, Prentice, & Mcilroy, 2019; Singer et al., 2016). However, this effect can be minimized through unpredictable timing of the cable's release. Furthermore, in the context of the present study, the practice effects and magnitude/direction predictability are less important due to participants having to recover from increasing lean angles/perturbation magnitudes up to their maximal ability. By increasing the perturbation magnitude and effectively increasing the difficulty of the balance response task until the point participants cannot recover balance, any practice effects and predictability are likely to be negligible on the kinetic and kinematic variables of their compensatory step restabilisation phase. That being said, the present study will utilise a maximum lean-and-release protocol to study the mechanics of the restabilisation phase as perturbation magnitude increases to the point that balance cannot be restored, which will aid in uncovering the mechanisms involved in a failed balance response.

Maximum Lean Angle Protocol

Several studies have utilized a maximum lean angle protocol to identify what factors are useful for effectively producing a compensatory step, how those factors change with increasing perturbations, and the differences between populations in the execution of a compensatory step. The maximum lean protocol can also be used to examine what occurs at the critical instance a person is unable to recover their balance, though very few studies have yet looked into this. This protocol is a modified version of the lean-and-release method in which the lean angle incrementally increases until the participant is unable to recover their balance with just a single step. In the literature it is generally accepted that young, healthy adults are most effective at performing compensatory stepping responses, and deviances from those patterns are likely due to challenges generating those responses. This line of thought is generally well placed though as examinations into the reactive stepping patterns of older adults reveals several challenges. In studies comparing older and younger adults, it is often found that young adults are able to recover from a significantly larger maximal lean angle; the maximal angles younger adults can recover from tend to range from 27.5 to 37.5 degrees whereas the older adults' lean angles range from 16.8 to 23.9 degrees (Hsiao-Wecksler & Robinovitch, 2007; Madigan & Lloyd, 2005; Thelen, Wojcik, Schultz, Ashton-Miller, & Alexander, 1997; Wojcik, Thelen, Schultz, Ashton-Miller, & Alexander, 1999). Therefore, examinations into the compensatory stepping patterns of young adults can help to reveal several strategies for producing an effective response.

Hsiao and Robinovitch (1999) identified three factors that, when coupled together, allowed for a successful compensatory step: step length, step execution time, and leg strength. This occurs as step length and leg strength increase and step execution time decreases. An increased step length creates a longer moment arm between the stepping foot and the CoM which consequently creates a greater decelerating moment acting about the CoM. Having a shorter step execution time will decrease the body's descent distance towards the ground, limiting the acceleration of the CoM and in turn decreasing the CoM's peak velocity. Having greater lower body muscle strength helps achieve these two other goals as increased force generation in the push-off will lead to greater swing velocity, consequently allowing for a faster execution time and a longer step length. Hsiao-Wecksler and Robinovitch (2007) further exemplified these ideas by testing female participants' (young and older adults) maximal ability to recover from a perturbation by altering their step length. The researchers measured the maximal lean angle that participants could recover their balance from using a single forward step; the length of this step either being 15%, 25%, 35% of their body height, or their maximal step length. As the step length increased, the young adults' maximum lean angle concurrently increased from 12.5° to 17.5° to 21.6° to 27.5°, respectively. The older adults experienced similar trends in the results with their maximum lean angle increasing from 10.3° to 13.5° to 14.3° to 16.8°, respectively. They also found there to be, for a given step length, a negative correlation between lean angle and step execution time in which larger perturbations require a faster stepping response to regain balance (Do et al., 1982; Hsiao-Wecksler & Robinovitch, 2007; Luchies et al., 1994; Maki, McIlroy, & Perry, 1996; Thelen et al., 1997). At a given lean angle the three factors of step length, step execution time, and leg strength work in conjunction to minimize the recovery effort of a compensatory step in the restabilisation phase by decreasing the necessary force output required of the stepping leg to halt the forward/downward CoM rotation. However, this recovery effort can increase (response becomes harder to accomplish) if reductions in one factor are not accounted for by enhancements in another (Hsiao & Robinovitch, 1999).

Though there are strategies one can implement to improve their compensatory stepping response, Hsiao-Wecksler and Robinovitch's (2007) results show that older adults are less able than young adults to recover from larger perturbations as exemplified in their maximum lean angles; young adults recovering balance when leaning forward up to 27.5° and older adults up to 16.8°. Other studies have found similar results with young adults being able to recover from a max lean of about 29.7° to 32.5° and older adults from a max lean of about 20.5° to 23.9° (Hsiao-Wecksler & Robinovitch, 2007; Madigan & Lloyd, 2005; Thelen et al., 1997; Wojcik et al., 1999). A few common results across these studies reveals that older adults, as compared to young adults, have a decreased maximal step length and longer step execution times in their respective maximal perturbations. The inability of older adults recovering from as great a lean angle may be attributed to the age-related decline in joint strength and flexibility as, at respective max lean angles, young adults use larger knee and hip ranges of motion (effectively producing a longer step) as well generate faster peak joint velocities in hip flexion, knee flexion and extension, and ankle plantarflexion (Madigan & Lloyd, 2005; Wojcik, Thelen, Schultz, Ashton-Miller, & Alexander, 2001). It should be noted though that as the lean angle increases, the step length and step execution time naturally become farther and faster, respectively, so some of these results may be due to older adults not needing as fast or long a step at their max lean angle compared to young adults at their significantly larger max lean angle (Do et al., 1982; Hsiao-Wecksler & Robinovitch, 2007; Luchies et al., 1994; Maki et al., 1996; Thelen et al., 1997). In addition, when comparing young and older adults at the same lean angle, there were no differences found in their range of motion or peak joint velocities (Madigan & Lloyd, 2005; Wojcik et al., 2001). It is possible that as the perturbation magnitude increases, older adults effectively reach their own limits of muscle strength and joint flexibility while young adults, at the same lean angle, have not. It is also possible that older adults, with increasing lean angles, become progressively more unstable following foot-contact to the point they are unable to regain stability. Despite the temporospatial and kinematic differences, the current literature regarding maximum lean angle studies has not yet delineated the specific kinetic biomechanical mechanisms limiting the older adults' ability to recover from larger lean angles. It is therefore critical to identify why older adults are not as able as young adults to recover from such large perturbation magnitudes to better understand the mechanisms of fall risk and to determine specific areas of focus for intervention.

Differences in the Compensatory Step Between Young and Older Adults

Much as the change-in-support strategy may be utilized in situations where a fixed-support strategy would suffice, older adults often employ multiple step strategies to recover from a loss of balance even when a single compensatory step would work (Luchies et al., 1994; McIlroy & Maki, 1996). Despite older adults not being able to recover from as great of lean angles as young adults, there are many similarities in the stepping patterns of both populations. In McIlroy and Maki's (1996) study, they sought to find age-related differences in compensatory stepping, in which they found both age groups to have very similar times to the onset of swing-leg unloading, to foot-off, and to foot-contact. In addition to this, both groups shared similar swing-phase durations, swing velocities, and changes to their CoM during the initial step. Though many of the factors from step-initiation to step-contact were very similar, they found one major difference between these two populations: the older adults were more likely to use multiple steps, as opposed to a single step, to regain their balance and stability (McIlroy & Maki, 1996). Even in studies with instructions to recover balance using only a single step, which has since been shown to have similar kinematics and kinetics to the initial step of a multi-step strategy (Cyr & Smeesters, 2007, 2009a), older adults are still more likely than young adults to take additional steps (Hsiao & Robinovitch, 2001). Hsiao and Robinovitch (2001), who instructed participants to take a single step following a backwards tether-release, further compared participants who primarily took multiple steps with participants who primarily took single steps and found there to be no difference in their step contact times, step length, contact force at foot-contact, or joint torques at foot-contact. A difference Hsiao and Robinovitch (2001) did find is that multi-steppers had a different body configuration at the instant of foot-contact in which they were leaning further backwards. While this may explain the necessity to take additional steps during backwards perturbations (i.e., a CoM closer to the posterior border of the BoS at foot-contact), it is difficult to extrapolate from these results if a similar issue persists in forward perturbations. The question then becomes, if the older adults are able to recover balance with a single step, why would a multiple step strategy be employed?

A common theory in response to this question is that the multiple step approach acts as a safety mechanism either by reducing the effort required to recover balance or by responding to instability that arises during the balance response task. To reduce the effort of a balance response task, older adults may initiate their step earlier than young adults as well as take multiple, shorter steps. When responding to large perturbations, a larger step and faster step execution time is required to regain balance (Hsiao-Wecksler & Robinovitch, 2007). When taking these larger and faster steps, a larger anteroposterior force impulse is generated following foot-contact that acts to slow, and eventually stop, the forward movement of the body and CoM. The issue here is that the larger anteroposterior impulses used to slow movement requires greater biomechanical strength (Won, 2001), and with age-related declines in muscle strength, older adults may be unable to produce these greater impulses. To counter this, older adults may take multiple short steps that have a reduced biomechanical demand (King et al., 2005). In this way, by minimizing the strength requirements of a balance-restoring task, older adults can ensure they regain balance even from larger perturbations despite the age-related consequences of reduced muscle strength. This is a stark contrast to young adults as they have been found not to minimize the strength requirements when producing a compensatory step (Hsiao-Wecksler & Robinovitch, 2007).

When tasked with stepping at 25% and 35% of their body height following a tether-release protocol, young adults were found to overshoot this distance (i.e., take a larger step than required) to a greater degree and more often than older adults, which may indicate that young adults do not minimize the biomechanical demand of their recovery efforts but rather ensure that they generate large enough stabilizing moments to not necessitate additional steps (Hsiao-Wecksler & Robinovitch, 2007). Young adults, however, do not face the same age-related challenges that older adults do (such as decreased muscle strength and flexibility), minimizing the younger population's need for additional steps. Contrary to some of this work, other research has shown that using either a single or multiple step response does not change the ability of young and older adults in recovering from an initial loss of balance (Cyr & Smeesters, 2007, 2009a). Cyr and Smeesters (2007, 2009a) examined the effect of using a single step, two steps, or an unlimited number of steps to recover balance from a maximum lean-and-release protocol. The researchers found that both young and older adults, within age-group, were able to successfully recover from a similar maximum lean angle and had no differences in the kinetic and kinematic measures of their initial step, regardless of how many steps the participants took (Cyr & Smeesters, 2007, 2009a). While the literature is divided in the usefulness of a multiple step approach as a way to reduce the biomechanical demand of a balance restoring task, extra steps have been shown to become necessary when instability arises during such a task.

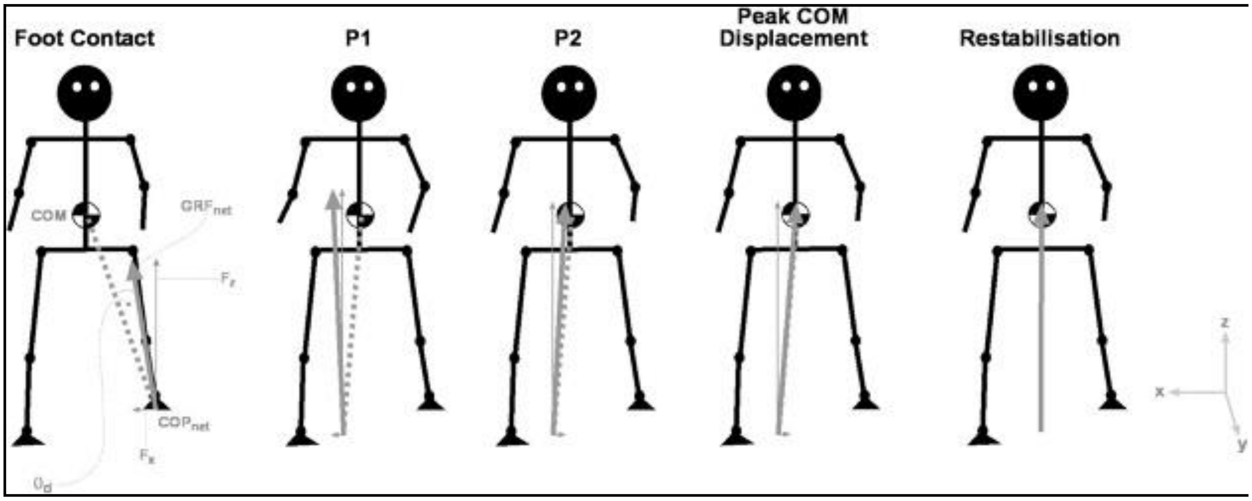
Luchies et al. (1994) has suggested that this multiple step approach is a conservative strategy that allows for more opportunities to correct for instability. McIlroy and Maki (1996) expanded on this idea when they compared the kinetic and kinematic measures between young and older adults following platform perturbations. They found that the multiple step strategy seen in older adults would often manifest as laterally directed steps following the initial step. These lateral steps were found to be a result of the presence of mediolateral instability that arose after foot-contact, which tended in the direction of the swing-leg side. The reactive lateral step is then, consequently, often taken in the direction of, and with, the swing-leg (McIlroy & Maki, 1996). At submaximal and expected perturbations, healthy older adults are generally able to recover their loss of balance using the multiple step approach, even in the presence of instability, though with a major caveat. With shorter initial steps (as compared to young adults), older adults have a more anterior position of their CoM relative to the anterior border of the BoS and they create smaller braking forces and stabilizing moments acting on/about their CoM, which may result in the necessity of additional steps to compensate (Arampatzis, Karamanidis, & Mademli, 2008; Karamanidis, Arampatzis, & Mademli, 2008; Maki, Perry, Nome, & Mcllroy, 1999). With the increased mediolateral instability following foot-contact seen in older adults (King, Akula, & Luchies, 2012; Singer et al., 2016), each additional step poses an extra risk of falling. So though additional steps may help in response to instability, each step poses its own risk of instability and a potential fall. Ultimately, the goal of a compensatory step is to prevent a fall from happening, however this is not always the outcome. A fall tends to occur not because of an inability to step, but rather due to an inability to restabilise. This means that if a person fails in decelerating and stopping their CoM through proper force production, a fall is likely to occur. The restabilisation phase is therefore especially important to examine.

The Restabilisation Phase

The restabilisation phase (Figure 3) occurs following foot-contact and is the only opportunity, where there is sufficient time, to generate sizeable restabilizing forces using both lower limbs. The restabilisation phase acts to minimize further displacements of the CoM (i.e.,

Figure 1.3

The Restabilisation Phase of a Forward Step



Note. From Singer, Prentice, & McIlroy (2016).

decelerate the CoM to stop within the BoS) to maintain dynamic stability. The CoM kinematics are controlled by modulating the net centre of pressure (CoPnet) as well as the magnitude and orientation of the net ground reaction force (GRFnet). Previous work has identified three discrete time points during restabilisation (foot-contact, P1, and P2), that occur with consistent timing, in which the GRFnet acts eccentric to the CoM – generating an external moment about the CoM – and alters the polarity of the CoM frontal and sagittal plane accelerations (Singer, McIlroy, & Prentice, 2014; Singer, Prentice, & McIlroy, 2019; Singer, Prentice, & McIlroy, 2013). The angle of divergence (θ_d), the angle made between the GRFnet vector and a line joining the CoPnet to the whole-body CoM, signifies the polarity, and is one factor determining the magnitude of the external moment acting about the CoM; a large and negative θ_d would create a large destabilizing moment whereas a small and positive θ_d would create a small restabilizing moment, respectively. As the external moment about the CoM is determined by both the magnitude and moment arm of the GRFnet, previous research has used θ_d, which disregards the magnitude of the GRFnet, as a means to specifically study age-related differences in the control of the GRFnet by the limbs (independent of the GRFnet magnitude).

During the swing phase, the GRFnet applies an eccentric force about the CoM, increasing its frontal and sagittal plane angular acceleration, and moving the CoM laterally (in the direction of the swing/stepping leg) and anteriorly, respectively. Due to the destabilizing external moment during the swing phase, the θ_d at the instance of foot-contact also acts as a destabilizing moment, accelerating the CoM towards the lateral (stepping leg side) and anterior borders of the BoS. The following two timepoints both represent the restabilizing effect of the GRFnet on the CoM kinematics: P1, the first restabilizing peak of the θ_d, occurs with consistent timing within 100ms following step contact and acts to slow the lateral and anterior movement of the CoM via acceleration in the opposite directions of the step. Due to its rapidity relative to foot-contact, however, P1 is believed to not be modulated by reactive control, but instead may be regulated proactively by preparatory muscle activity and limb stiffness on foot-contact. P2, the second restabilizing peak of the θ_d, occurs about 250 ms following foot contact - between P1 and the eventual restabilisation of the CoM. P2, directed in the same way as P1, also acts to further slow the displacement of the CoM. In addition, because of its temporal properties, P2 may be modulated reactively in response to the CoM kinematics following step contact, which may allow it to scale with the amount of instability present (Singer et al., 2019, 2016). Previous work has found moderately high correlations between the timing of P2 relative to foot-contact and the extent of instability occurring during restabilisation (Singer et al., 2016). Because of this, the characteristics of P2 may be critical components of the restabilisation phase. Shortly following P2, the CoM reaches its peak displacement before reversing directions and moving medially and posteriorly until the eventual point of restabilisation.

Several studies have shown that older adults, during the restabilisation phase, have difficulties in controlling their mediolateral (ML) and anteroposterior (AP) dynamic stability (Singer et al., 2014, 2019, 2013, 2016). There are a few key ways in measuring one's instability, including the CoM incongruity and trial-to-trial variability. For the ML and AP direction, the CoM incongruity is the difference between a participant's peak lateral or anterior CoM displacement, respectively, and the CoM position at the point of restabilisation. Large incongruity values mean the CoM travels a greater distance and moves closer to an individual's stability limits, suggesting difficulties in decelerating the CoM to stay within the BoS. Trial-to-trial variability is the measure of the variance of the CoM incongruity between trials. Increases in this variability suggests difficulties in the ability to consistently control the CoM kinematics, as well as the increased probability of generating a large CoM incongruity following foot contact (Singer, 2016). Though a person can regain stability despite having a large CoM incongruity value (i.e., the CoM reaches close to but never extends beyond a person's stability limits), a large variability of this measure indicates that the CoM, over the course of several balance response tasks, is more likely to extend beyond this threshold. In other words, while the majority of responses may result in balance and stability being achieved, an increased variability will subsequently increase the likelihood of one of many responses resulting in a fall.

In Singer et al. (2013), older and younger adults performed multiple voluntary single-step tasks at various speeds and foot placements. The researchers found the older adults to have a reduced magnitude of both P1 and P2 in the ML direction, effectively limiting the net external stabilizing moment acting about the CoM. However, this was likely a result of the reduced effort required by the older adults to restabilize due to strategies implemented in the step initiation and swing phases (i.e., prolonged anticipatory postural adjustments) to reduce their frontal plane

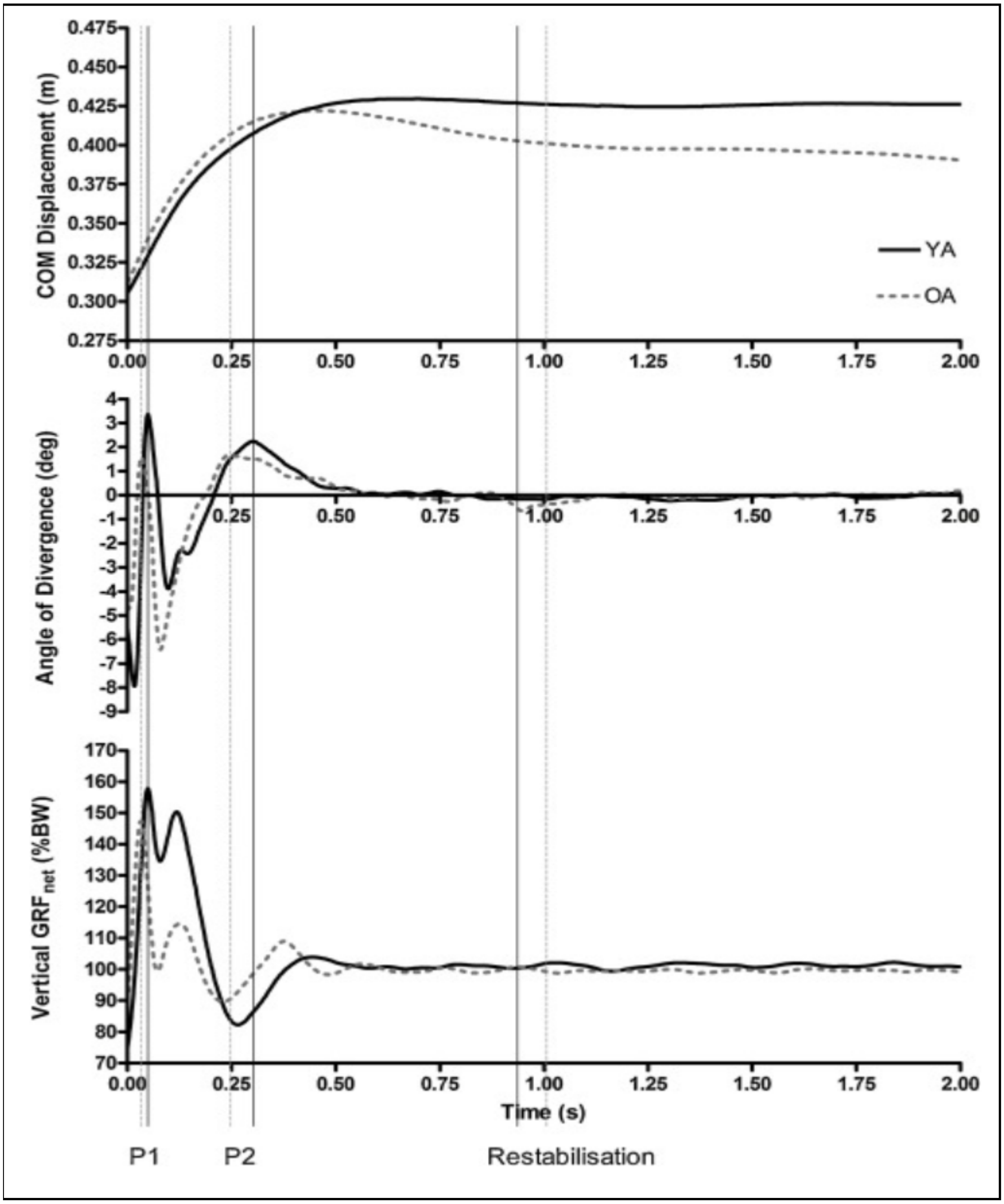
CoM acceleration. Despite these pre-emptive strategies, which should have increased ML stability during the restabilisation phase, the older adults were still found to have an increased ML CoM incongruity and trial-to-trial variability during restabilisation when compared to the young adults (Singer et al., 2013). Due to the age-related stability issues seen in voluntary stepping, especially with the presence of a lengthened anticipatory postural adjustment, it begs the question of what challenges older adults face during a compensatory stepping response following an unknown perturbation.

In a later study from Singer et al. (2016), the relationship between the whole body CoM and the orientation of the GRFnet was examined to explore the control of dynamic stability during the restabilisation phase following unpredictable perturbations, with special attention towards the age-related challenges of ML stability control. Using a lean-and-release protocol with a pseudorandom time delay for initiating the perturbation, it was found that older adults, compared to young adults, had a greater ML CoM incongruity, greater intertrial variability of the ML CoM incongruity, and an increased time to P2 (Singer et al., 2016). However, with only two blocks of five trials, the ML CoM incongruity, intertrial variability, and the magnitude of the θ_d at foot-contact (the magnitude of the destabilizing moment at foot-contact) were reduced in the second block of trials suggesting potential learning effects from the participants. They also found that young and older adults, between-groups, had similar magnitudes of the θ_d at P1 and P2. Lastly, the researchers found a significant positive correlation between the older adults ML CoM incongruity measures and the timing of P2, suggesting that the age-related challenges in ML stability control may be linked to the reactive control of the applied force (i.e., the timing and orientation of the GRFnet) during the restabilisation phase (Singer et al., 2016).

While these former studies focused on the ML instability, a more recent paper by Singer et al. (2019), using a similar methodology, examines the relationship between the CoM and GRFnet though with an emphasis on AP stability control (Figure 4). Similar to the ML direction, the researchers found the older adults to have a greater AP CoM incongruity magnitude and variability in the restabilisation phase as compared to young adults. Unlike the ML direction, the were no differences between populations in the magnitude or timing of P2, however, older adults did exhibit a reduced magnitude of the AP GRFnet eccentricity at P1. In addition, the researchers found there to be a positive correlation between the AP CoM incongruity magnitude and the P1 magnitude (Singer et al., 2019). A study from Karamanidis, Arampatzis, and Mademli (2008) shares similar results as they had young and older adults recover from two different lean angles and examined their kinetic, kinematic, and musculotendinous properties during a compensatory step. They found that young adults, following foot-contact, produced larger braking AP GRFs that helped to slow the anterior movement of the CoM, and this being largely due to greater knee extensor muscle strength and stiffness that then allows for producing and recovering from a longer forward step (Karamanidis, Arampatzis, & Mademli, 2008). Together, this suggests that regulating anterior instability may take priority through the generation of a large P1 eccentricity via proactive control of limb stiffness prior to step contact. Lateral instability may be managed reactively following foot contact, via the P2 eccentricity, in response to the residual mediolateral instability. This recent research has led to new insights on the challenges faced by older adults and the biomechanical mechanisms related to ML and AP instability during the restabilisation phase. This research further highlights that the issues older adults face in dynamic stability are present even in fairly simple, common, day-to-day tasks. However, previous research has not examined how the restabilisation phase is impacted by perturbations of increasing magnitudes;

Figure 1.4

AP Measures of the Restabilisation Phase of a Compensatory Step



Note. Figure 4 shows a representative trial from an older adult (OA – grey dashed) and a younger adult (YA – black solid). Foot-contact occurs at Time=0s. Top: CoM Displacement (m) relative to the initial CoM position, prior to perturbation. Centre: Angle of Divergence (deg). Bottom: Vertical component of the GRFnet, expressed as a percentage of body weight (%BW). From Singer, Prentice, & McIlroy (2019).

with restabilisation being critical to maintain stability and prevent a fall, research has also not examined what happens when the restabilisation phase fails to restabilize.

Instability During Restabilisation

While many of the mechanisms that induce a state of instability are known, as well as many factors that lead to an increased risk of falling during a compensatory step are known, the specific kinetic mechanisms that occur when a compensatory stepping task fails are not well understood. It has been shown that the restabilisation phase is where the body will either become stable (successful balance recovery) or remain unstable (failed balance recovery), and this being largely dependent on the magnitude and orientation of the GRFnet and its effects on CoM kinematics. Previous research has revealed age-related differences in the restabilisation phase of a compensatory step - in the ML direction these include older adults having a greater ML CoM incongruity (indicating instability), an increased trial-to-trial variability of their ML CoM incongruity, an increased time to P2, as well as a strong correlation between ML CoM incongruity measures and the timing to P2 (Singer et al., 2016). In the AP direction, these age-related differences include older adults having a greater AP CoM incongruity, an increased trial-to-trial variability of their AP CoM incongruity, a decreased magnitude of the GRFnet eccentricity at P1, as well as a strong correlation between AP CoM incongruity and the decreased magnitude at P1 (Singer et al., 2019). What this boils down to is an impaired ability for older adults in controlling their dynamic stability following foot-contact via challenges in producing proper magnitudes and orientations of their GRFnet. However, these results come from studies utilizing a submaximal perturbation that is generally easy for healthy young and older adults to perform safely. While the presence of AP and ML instability in such tasks is concerning and should be a target of fall-related interventions, healthy older adults who have presented with this instability are still well able to regain their balance and stability at low perturbation magnitudes. Although, outside of laboratory environments, in general instances a person must use a compensatory step to avoid falling, the perturbation magnitude cannot be controlled; as the magnitude increases, so too will the risk of falling.

In the literature regarding maximum lean-angle studies, two common streams of thought for why older adults are unable to recover from perturbation magnitudes as large as young adults include: (1) that the age-related decline in joint strength and flexibility limits the older adults' ability to make a large, fast enough step and (2) that older adults face an increasing amount of instability following foot-contact as the lean angle increases to the point they cannot restabilise themselves. While the answer is likely a mix of both with one factor impacting the other, it is unknown from the current literature how debilitating each factor is in the execution of a compensatory step. In other words, it is unknown how these restabilisation phase variables change as the perturbation magnitude steadily increases, as the subsequent balance response task becomes more difficult, and in instances someone is unable to regain their balance with just a single step. With increasing perturbations, does the issue lie in an inability to properly scale and adapt the GRFnet magnitude, the GRFnet orientation, or both? And if both, to what degree does each impact the increasingly difficult compensatory step? With a better understanding of the mechanics of the restabilisation phase at varying perturbation magnitudes and in the instance of a failed compensatory step, interventions can better target such mechanisms to reduce patients' risk of falling.

Purpose

The purpose of this research is to examine the relationship between the whole-body center of mass and the magnitude/orientation of the net ground reaction force during the restabilisation phase of compensatory stepping as the perturbation magnitude increases until balance cannot be achieved with a single step. By utilizing a maximum lean-and-release protocol, this study will explore how this relationship changes and/or adapts (1) as perturbation magnitude increases and (2) at the critical instance where the balance response task fails. In other words, the purpose of this study is to (1) examine how people adjust their restabilisation response to successfully recover from increasing instability, and (2) examine the limiting factor(s) that prevents people from recovering balance when they experience too much instability. While older adults are shown to have instability in the restabilisation phase following submaximal perturbations, this initial exploratory work will focus on young adults. The changes in the young adults' restabilisation phase mechanics with increasing perturbations, which should result in increasing instability, will be compared to the instability issues seen in older adults in future studies. This will allow the focus of the research to be contained to the mechanisms governing whether restabilisation is achieved, while still being able to understand and contrast what instability issues will occur.

Hypotheses

Successful Balance Recovery Responses with Increasing Perturbation Magnitudes

It is hypothesized that for successful balance recovery responses as the perturbation magnitude increases, participants will experience increasing instability at and following foot-contact that will necessitate a greater restabilisation response to recover from. More specifically, it is hypothesized that an increasing lean angle will result in a decreased AP MoS and larger AP destabilizing moment at the instance of foot-contact which will require a longer step length, larger AP stabilizing moments at P1 and P2, and a shorter time to P1 and P2 so as to prevent the CoM from extending beyond the anterior boundaries of the BoS. As the increasing lean angle would result in an increased temporal urgency to initiate the compensatory stepping response and thus reduce the potential of producing an effective ML anticipatory postural adjustment, it is also hypothesized that there will be a decreased ML MoS and increased ML destabilizing moment at the instance of foot-contact. However, as there is greater necessity to stabilize the AP direction first, it is further hypothesized that the step width, ML stabilizing moments at P1, and the time to P1 will not significantly change. Due to the hypotheses of decreased ML stability at foot-contact but no difference in the initial ML response, it is hypothesized that the stabilizing moment at P2 will significantly increase and the time to P2 will significantly decrease (as P2 may be modulated reactively in response to instability).

Failed Balance Recovery Responses

For when the balance response task fails to restabilize, it is hypothesized that this will occur in one of two different scenarios; (1) due to the XCoM at the moment of foot-contact being outside the borders of the BoS (negative MoS) or (2) due to an inability to produce large enough stabilizing moments that act to slow the movement of the CoM. In the first situation, this is hypothesized to occur primarily in the AP direction as a result of the step length being too short at foot-contact, given the increased CoM velocity. For the second situation, this is hypothesized to occur in the AP or ML direction as a result of large destabilizing moments at foot-contact, decreased or insufficient stabilizing moments P1, as well as an increased time to, or absence of, P2. This study also explores what variables may predict whether a balance response task is more likely to succeed or fail. It was hypothesized that the predictors of a failed balance response task (i.e., what increases the likelihood of a balance response task failing) include a decreased step length and a decreased (more negative) AP θ_d at foot-contact.

Chapter II: Study Design

Participants

This study recruited 25 healthy young adults aged 19-32 years old (24.84 ± 4.22 years). Sample size estimates were obtained using the smallest effect size (f=0.176) obtained from previous work examining differences found in young adults' GRFnet eccentricity at P1 with trial-repetition during a lean-and-release protocol (Singer et al., 2019), along with $\alpha = 0.05$ and $\beta = 0.80$. GRFnet eccentricity was chosen as it is one of the key outcome measures in this study, while the effect of trial-repetition was chosen as it was the closest analogy for examining the kinetic measures of the restabilisation phase with increasing perturbation magnitudes. The exclusion criteria involved the use of the Physical Activity Readiness Questionnaire for Everyone (PAR-Q+); a participant would have been excluded from the study if they indicated "Yes" on any of the general health questions, which ensured participants were able to participate in physical activity, they had no injuries/conditions that could be made worse or inhibit them from participating in this study, and that they had no balance-related issues, among others. Additionally, participants would have been excluded from this study if they indicated they had an allergy to latex, as the tape used to affix retroreflective markers to participants could be an irritant. All participants were required to be able to stand, walk, and perform a sub-maximal compensatory step unaided. It was also required to obtain informed written consent from each participant prior to data collection.

Instrumentation

Eight Vicon Vero cameras (model 2.2, Vicon Motion Systems, Los Angeles, CA, USA) were used to record kinematic data at 100Hz while four force platforms (two are model 9260AA6, the other two are model 9286BA, Kistler, Amherst, NY, USA) were used to record ground reaction forces and moments at 2000Hz. While these sampling rates are in excess of the minimum sampling frequency defined by the Nyquist Sampling Theorem, force data were oversampled to ensure we could precisely detect the onset of the perturbation from the force transducer; individual marker data were oversampled to ensure there was no blurring of markers, which could affect marker centroid calculations. The force platforms were arranged in a T-shaped array where participants stood with their feet side-by-side on two separate force platforms and stepped forward onto a third, then fourth, force platform. All of the motion capture data and analogue-to-digital signals were synchronously recorded by Vicon Nexus software (version 2.7.0, Vicon Motion Systems, Los Angeles, CA, USA).

Participants were equipped with retroreflective calibration and tracking markers (1cm in diameter) placed bilaterally at anatomical landmarks consistent with previous work (Singer, Prentice, & McIlroy, 2012). The calibration markers defined the segment endpoints for the head, trunk, pelvis, and upper and lower limbs; they allowed for the determination of segment geometry, the inertial properties, CoM locations, and local coordinate systems of each segment. The calibration markers were worn by the participants during the standing reference trials, then subsequently removed. The tracking markers were worn by the participants during the standing reference trials as well as the data collection trials. Following the standing calibration, tracking markers had a known position within each segment coordinate system and were used to track the position and orientation of each segment.

The calibration markers were placed on the sternum and xiphoid process, and bilaterally on the greater tubercle of the humerus, anterior to external auditory meatus of the ear, medial and lateral humeral epicondyle, radial and ulnar styloid processes of the wrist, head of the 3rd metacarpal, anterior superior iliac spine, medial femoral condyle, medial malleolus of the tibia, medial aspect of the calcaneus, and the base of the 1st metatarsal.

The tracking markers were placed bilaterally on the acromion process, iliac crest, greater trochanter of the femur, lateral femoral condyle, lateral malleolus of the fibula, distal phalanx of the great toe, head of the 5th metatarsal, lateral aspect of the calcaneus, and the supralateral aspect of the forefoot. There were also rigid clusters of 4 markers placed bilaterally on the thigh, shank, upper arm, lower arm, sacrum, and the upper back between the scapulae. A head strap consisting of 8 markers was worn around the head at the height of the forehead.

The individual retroreflective markers were placed on the participants' skin/clothing over the specified landmarks using double-sided tape. The marker clusters and head strap, with the exception of the upper and lower back clusters, were affixed directly over the participants' skin using nonslip bands secured with Velcro tabs. The upper and lower back clusters were affixed snugly over the participants' shirt using Velcro straps.

Positive x-, y-, and z-axes for the laboratory coordinate system were oriented to the right of the participant, anteriorly, and superiorly, respectively. In the cases where participants stepped forward with their left leg, the polarity of the frontal plane data was reversed, to maintain consistency with right-footed stepping trials.

Protocol

At the beginning of the data collection session the participants were given a consent form to sign. Participants then completed the Physical Activity Readiness Questionnaire for Everyone (Par-Q+), the Physical Activity and Sedentary Behavior Questionnaire (PASB-Q), the Falls Efficacy Scale – International (FES-I), and a demographic questionnaire collecting the participants' height, weight, sex, and age.

The participants were then outfitted with the calibration and tracking retroreflective markers. Once equipped, the participants performed a static calibration trial which allowed for the determination of segment anthropometry and the determination of local, segment embedded, coordinate systems for each segment. Following this, the calibration markers were taken off of the participants while leaving the tracking markers on. The participants then performed four quiet standing reference trials which involved standing on two force platforms (one platform per foot) for 60 seconds. The two initial standing reference trials had their feet side-by-side in a standardized position, as per McIlory and Maki (1997) – the heels 17cm apart from one another with each foot at a 14° angle between the long axes of the feet. The next two reference trials used a terminal stance configuration where their feet were staggered with one foot in front of the other, with each trial having a different forward foot. Altogether, these trials mimicked the initial and final stances that were seen in the experimental trials and were used to obtain the baseline kinematic variables necessary to determine the initial and final conditions of the participants' compensatory step. Following this, the participants performed the Modified Clinical Test of Sensory Interaction in Balance (CTSIB-M) which involved four 30 second standing balance trials – with feet side-by-side – consisting of the participant with (1) their eyes open standing on a firm surface, (2) their eyes closed standing on a firm surface, (3) their eyes open standing on a foam pad, and (4) their eyes closed standing on a foam pad. This test assesses the ability of an adult in using sensory inputs when one or more sensory systems are compromised (i.e., vision is removed in trials 2 and 4, the somatosensory system is removed in trials 3 and 4, and the vestibular system is used in all trials and is the only system used in trial 4).

For the experimental trials, participants performed multiple perturbation-evoked compensatory stepping tasks via the lean-and-release protocol. This involved strapping participants into a harness with a superior connection attached to a ceiling-mounted rail system via a rope of adjustable length (to prevent participants from contacting the floor should a fall occur) and a posterior connection attached to a detachable cable (to allow for a forward lean and to invoke a perturbation). The rope of the superior connection was adjusted to a length that prevented the participants' hands from being able to touch the ground while they were bending forward with their legs kept straight; this length stops participants from hitting the ground in the event of a fall. The posterior cable allowed the participants to lean forward into the harness without falling until the cable was detached from its anchor point, inducing a forward perturbation. Participants began each trial with their feet side-by-side with each foot on a separate force platform. Leaning forward, the participants were required to have their mediolateral weight equally distributed under each foot (within 10%). This ensured participants were not preloading their stance limb prior to step initiation, which could reduce the time to step initiation and decrease consequent mediolateral instability developing during swing phase, by limiting lateral CoM accelerations. Further, participants were required to have a consistent anteroposterior position of their CoP under each foot for each lean angle, which ensured individuals were not differentially preactivating the ankle plantarflexors across conditions, prior to perturbation onset. Upon reaching these starting conditions, the posterior cable was detached following a pseudo-random time delay to ensure the participants could not predict when the perturbation would occur. Participants were asked to recover their balance with a single step of self-selected step characteristics, and to maintain their new stance for approximately 10 seconds after restabilisation to establish final conditions. After these 10 seconds, the trial was considered complete.

A trial was either considered to be a "success" or a "failure". A failed trial occurred when a participant was either (1) unable to recover their balance or restabilise with just a single step (i.e., using multiple steps or falling) or (2) when 20% of their bodyweight was being supported by the superior connection of the harness (Cyr & Smeesters, 2009b). A successful trial occurred when a participant recovered their balance and restabilised with a single step, and if less than 20% of their bodyweight was supported by the superior connection of the harness.

As measured by a force transducer on the detachable cable, participants initially leaned forward with the posterior cable supporting 10% of their body weight. This equates to a lean angle that, when the cable is released, requires a compensatory step to recover balance (Singer et al., 2016). After 3 successful trials, the lean angle was increased by having the posterior cable support an additional 5% of their body weight when leaning forward (i.e., 15% of their bodyweight was being supported by the posterior cable). After every 3 successful trials, the lean angle was further increased by having another 5% of their bodyweight being supported by the posterior cable. This process was repeated until a participant had 2 failed trials at a given lean angle. Participants were given at least 1 minute of rest between lean angle conditions to reduce the effect of fatigue throughout the data collection session. The largest lean angle in which a participant was able to successfully restabilise for 3 trials, without achieving 2 failed trials, is considered to be their Maximum Lean Angle. The lean angle in which the participant achieved 2 failed trials is considered to be their Failed Lean Angle. Upon reaching a participant's Failed Lean Angle, or if a participant indicated they did not want to continue trials, the data collection session was considered complete.

Data Processing

All motion files were processed using Visual 3D software. Interpolation of missing marker trajectories was performed using quintic splines (Howarth & Callaghan, 2010; Wood & Jennings, 1979). Marker interpolation was used to fill any gaps of the marker data up to 30 frames (0.300s) in length. For the force platform data, a lowpass, zero-lag, 4^{th} order, Butterworth filter was used with a cut off frequency of 15Hz. For the individual marker data, a 20^{th} order, critically damped filter was used with a cut off frequency of 6Hz. The critically damped filter was used to ensure good time domain properties for kinematic data; the filter order was increased to 20 (5 bidirectional passes) to ensure equivalent frequency domain properties to the Butterworth (Robertson & Dowling, 2003). These filters work to attenuate high-frequency noise caused by errors in the automatic processing of marker locations and in the vibrations recorded by the force platform.

Data Analysis

Successful trials contain kinematic and kinetic data ranging from when the initial conditions of the trials were met up until 10 seconds following foot contact, which exceeds the time required for restabilisation. The data used in statistical analyses for the successful trials were limited to the range from perturbation onset (the release of the cable) to the point of restabilisation. Failed trials were also ended 10 seconds after the initial foot-contact, however the data used in statistical analyses for the failed trials were limited to the range from perturbation onset to the point of failure. The point of failure was determined based on when the participant was considered to have failed that trial. In the case a participant failed a trial due to taking a second step, the point of failure was considered to occur when the foot making the second step had a vertical ground reaction force of less than 10.00 N (i.e., when the second stepping foot left the ground). In the case a participant failed a trial due to the harness' superior connection supporting at least 20% of the participant's bodyweight, the point of failure was considered to occur when the force transducer measuring this value exceeded the 20% threshold. In the case a participant failed a trial due to both of these conditions, the point of failure was considered to occur based on which event happened earliest. Data after this point were not considered as this study aims to assess why the first step was not able to recover balance; if taking a second step, then the measures after the second foot leaves the ground would not be accurate to the participants' first step; if supported by the harness, the influence of the harness physically stopping or slowing the participants' movement would cause measures after this point to not be accurate to the participants' natural response.

To determine when certain events occurred, the following measurements and calculations were performed. Perturbation onset was determined to occur at the instant the force transducer measuring the posterior cable load dropped from its steady-state maximal value. As the transducer timeseries resembled a descending step function, with the steady-state maximum value given by the percentage of the participant's body weight in a given condition, the threshold to determine perturbation onset was specific to each participant and condition. The most efficient method to determine perturbation onset for all participants and conditions was to differentiate the step function, which resulted in an impulse function – perturbation onset was determined to be the first instant where the value deviated from zero, prior to the peak of the impulse. The time to onset of ML asymmetry was determined as the instant at which the vertical force (Fz) signals from each limb differed by 2% of body weight (McIlroy and Maki, 1996).

Toe-off of the stepping limb was determined to occur at the first sample of data in which the stepping foot's vertical ground reaction force was less than 10.00 N; the time to step onset was determined to be the time from perturbation onset until toe-off. Foot-contact was determined to occur at the first sample of data in which the stepping foot's vertical ground reaction force was greater than 10.00 N; the swing time was determined to be the time from toe-off until foot-contact. Restabilisation was considered to occur when a participant's anteroposterior or mediolateral shear component of their GRFnet entered and remained within a 2 SD bandwidth determined by the last 2 seconds of the respective trial (for anteroposterior and mediolateral measures, respectively); this ensures the horizontal CoM acceleration is, or is close to, 0 m/s^2 as there can only be a shear component to the GRFnet if the CoM acceleration is greater than 0 m/s^2 (assuming no external forces are acting on the body, besides gravity). The time to restabilisation was determined to be the time from foot-contact until restabilisation.

Temporospatial data of the stepping foot, such as the step length and step width, were determined by calculating the respective AP and ML distances between the stepping and stance limb great toe and lateral calcaneus markers, respectively, between toe-off and foot-contact. Step length and width were normalized to body height.

The GRF values from the multiple force platforms were combined to determine the GRF_{net} and CoP_{net}. Time series data were calculated, with respect to the AP and ML axes of the global coordinate system, from the sagittal and frontal plane inclination angles of (1) the GRF_{net}

vector and (2) a line joining the CoP_{net} and CoM. The angle of divergence (θ_d) was then calculated as the difference between these two waveforms (see Figure 4); this measure signifies the eccentricity of the GRF_{net} relative to the CoM (Singer et al., 2014). A negative θ_d represents a destabilizing effect of the GRF_{net} on the CoM while a positive θ_d represents a stabilizing effect. Two time points along the θ_d waveform were identified, including (1) the first positive peak (P1) being the maximum θ_d magnitude that occurred within the 100ms following foot-contact, and (2) the second positive peak (P2) being the maximum θ_d magnitude that occurred between the 100ms following foot-contact and restabilisation. The θ_d magnitude in the AP and ML directions were recorded at foot-contact, P1 (termed 'AP P1' and 'ML P1', respectively), and P2 (termed 'AP P2' and 'ML P2', respectively), as well as the timing of these events.

Time series data of the GRFnet were used to determine the AP GRF and the ML GRF magnitudes, which act in the sagittal and frontal planes, respectively. As the eccentricity of the GRFnet is a proxy measure of the moment arm of the GRF, we also sought to examine the magnitudes of the GRFnet components acting in each the sagittal and frontal planes, to understand whether challenges in restabilisation could result from insufficient force production (in addition to improper directionality in force production. From the GRFnet data, the sagittal plane (AP) GRF component was determined by combining (by vector addition) the anteroposterior horizontal force and the vertical force components into a single waveform, while the frontal plane (ML) GRF component was determined by combining the mediolateral horizontal force and the vertical force into a single waveform. Timeseries data of the AP and ML GRF were collected as well as the specific values at perturbation onset, toe-off, foot-contact, and the respective AP and ML P1 and P2.

Further measures of AP and ML instability, as well as balance was determined by the Margin of Stability (MoS). The MoS defines the relationship between the anterior/lateral limits of the BoS and the velocity-adjusted, or extrapolated, position of the center of mass (X-CoM) (Hof, Gazendam, & Sinke, 2005). As a compensatory step is a dynamic movement with the goal of restoring balance, a large CoM velocity must be slowed to remain within the boundaries of the new BoS. As the name suggests, the X-CoM is a measure that adjusts the horizontal position of the CoM based on the magnitude and direction of the CoM velocity; the larger the velocity, the further the X-CoM is adjusted relative to its spatial position. This can result in situations where the CoM is close to, but within, the stability limits of the BoS though due to a high velocity of the CoM directed outward, the X-CoM is considered to be outside the BoS. The AP and ML X-CoM was then compared to the anterior and lateral limits of the BoS, respectively, to determine the respective AP and ML MoS. A small MoS value (i.e., closer to 0) indicates that the X-CoM is close to the limits of the BoS, while a larger MoS value (i.e., further from 0) indicates that the X-CoM is further from the limits of the BoS. A positive MoS value indicates that the X-CoM is within the borders of the BoS and thus the body is stable in its current state, while a negative MoS value indicates that the X-CoM is outside the borders of the BoS and thus the body is unstable in its current state, requiring the generation of a large restorative force or a change in the BoS configuration. The most stable MoS value would then be a large, positive value (Hof et al., 2005). Timeseries data of the MoS was collected as well as the specific AP and ML MoS values at perturbation onset, toe-off, foot-contact, and the respective AP and ML P1 and P2.

In summary, the identified events include the perturbation onset, toe-off of the stepping limb, foot-contact of the stepping limb, P1 in the AP and ML directions, P2 in the AP and ML directions, and restabilisation. The dependent variables include the time to step onset, time to onset of ML asymmetry, swing time, step length, step width, the AP and ML MoS (at perturbation onset, toe-off, foot-contact, and the respective AP and ML P1 and P2), the AP and ML θ_d (at foot-contact and the respective AP and ML P1 and P2), the AP and ML GRF (at foot-contact and the respective AP and ML P1 and P2), the time to AP and ML P1 and P2 (relative to foot-contact), and the time to restabilisation.

Statistical Analysis

Analysis of Successful Trials with Increasing Lean Magnitudes

Paired samples T-tests were utilized to understand the effect of an increasing perturbation magnitude on the balance response task by comparing participants' initial lean angle and maximal lean angle. The paired samples T-test were run on the dependent variables found in Table 3.1, comparing the initial lean angle to the maximal lean angle with a significance level of $p < .05$.

The initial lean angle for all participants was 10%BW (N=25, 100%), while the maximal lean angle included 25%BW (N=8, 32%), 30%BW (N=8, 32%), 35%BW (N=8, 32%), and 40%BW (N=1, 4%); the difference in maximal lean angles is due to participants having different lean angles in which they achieved 2 failed trials (Table 3.2). The participants' 3 successful trials for each successful lean magnitude were averaged within subject.

Analysis of Failed Trials Compared to Successful Trials

Independent samples T-tests, Pearson correlations, and binary logistic regressions were performed to understand if there were significant predictors of a successful or failed balance recovery response. These tests compared successful and failed trials that occurred at the same lean angle which was important for comparison's sake in establishing the initial conditions of the perturbation to be similar between successful and failed trials; this would mean that a participant failed due to their individual response to the perturbation as opposed to their perturbation being larger than the successful group's perturbation.

All participants (N = 25, 100%) participated in trials at a lean angle of 30% BW; seventeen of those participants were successful (N = 17, 68%; they achieved 3 successful trials and ≤ 1 failed trial) and the remaining eight participants failed (N=8, 32%; they had 2 failed trials and ≤2 successful trials). At a lean angle of 35%BW, of those previously successful seventeen participants (N = 17, 100%), nine participants were successful (N=9, 53%) while eight participants failed (N = 8, 47%). At a lean angle of 40%, of those previously successful nine participants (N = 9, 100%), one participant was successful (N=1, 11%) while eight participants failed (N = 8, 89%). At a lean angle of 45%, the one remaining participant failed (N=1, 100%). Because the lean angles of 30%BW and 35%BW had the largest sample sizes (N = 25, N = 17, respectively) with the most even split between successful and failed trials (successful:failed; 17:8, 9:8, respectively), the lean angles of 30%BW and 35%BW were chosen to be included in the analysis while 40%BW (N = 9, 1:8) and 45%BW (N=1, 0:1) were excluded from the analysis. There were two separate analyses: one for the 30%BW lean angle trials, and the other for the 35%BW lean angle trials. The measures for a participant's 3 successful trials were averaged within-subject for each successful lean magnitude and a participant's 2 failed trials at their failed lean magnitude were averaged within-subject.

Independent samples T-tests were performed to determine the differences between the successful and failed balance recovery responses for each of the following variables: AP and ML MOS at perturbation onset, time to onset of ML asymmetry, time to step onset, AP and ML MOS at toe-off, swing time, step length, step width, AP and ML MOS at foot-contact, AP and ML θ_d at foot-contact, and AP and ML GRF at foot-contact. To ensure the initial conditions between the successful and failed groups were the same at the same lean angle, there needed to be no significant differences for the AP MoS at perturbation onset and the ML MoS at perturbation onset. A significance level of $p < .05$ was used.

The variables measured at events following foot-contact (i.e., P1, P2, and restabilisation) were not included in this analysis as most failed trials had their point of failure occur prior to these events and thus could not be calculated, though there were some exceptions. At the 30%BW lean angle, only 1 of the 8 failed participants was missing the AP and ML P1 data, so a separate analysis was run for the 7 failed participants who did have P1 data. Of those 7 participants, 5 participants had an AP P2 response and, of those 5, 3 participants had an ML P2 response; the P2 data was deemed to be from too small a sample to analyse statistically. At the 35%BW lean angle, all participants had ML P1 data, so this was included in the analysis. Of the 8 participants with failed trials at the 35%BW lean angle, 4 had an AP P1 response and no participants with failed trials had an AP or ML P2 response; this was deemed to be too small a sample to analyse statistically.

The results of the independent samples T-tests were also used to inform which variables would be considered to include in the binary logistic regression. For the logistic regression, the dependent variable was the balance response outcome (successful = 0, failed = 1) while the independent variables were chosen from the measures examined in the T-tests. To fit with the logistic regression's assumption that the independent variables be linearly related to the logit transformation of the dependent variable, only measures with a significance value of $p < .200$ from the T-tests were considered to be included in the logistic regression models as predictors of a failed balance response.

Pearson correlations were then performed on the considered predictors to determine which variables would be included in the binary logistic regression, with a significance of $p < .05$ being used. To fit with the logistic regression's assumption that there should be no multicollinearity among the independent variables, variables that were significantly correlated with one another were not included in the same model. With the assumptions of the binary logistic regression model met, the predictor variables of each model were determined.

Separate binary logistic regressions were performed for the lean angles of 30%BW and 35%BW, as well as for the AP and ML measures (Table 2.1). For the 30%BW AP logistic regression, the variables of time to step onset and step length were included in the model. For the 30%BW ML logistic regression, the variables of time to step onset, ML MoS at toe-off, and step width were included in the model. For the 35%BW AP logistic regression, the variables of time to step onset, step length, and AP θ_d at foot-contact were included in the model. The 35%BW ML logistic regression could not be run as only 1 dependent variable fit the criteria for being included in the model.

The binary logistic regressions determined the likelihood that an independent change in a predictor's value would result in a failed balance recovery response, otherwise known as the odds ratio. Odds ratios that are > 1 indicate that a failed trial is more likely to occur due to an increase in the predictor's value. Odds ratios that are < 1 indicate that a failed trial is less likely to occur due to a 1.0 value increase in the predictor's value. 95% Confidence Intervals (95%CI) will be used to determine significance.

Table 2.1

Predictors in the Binary Logistic Regression Models

30% BW		35% BW	
Anteroposterior	Mediolateral	Anteroposterior	Mediolateral
Time to Step Onset	Time to Step Onset	Time to Step Onset	
Step Length	ML MoS at Toe-Off	Step Length	
	Step Width	AP θ_d at Foot-Contact	

Chapter III: Results

Testing was completed with twenty-five participants. The sample included thirteen females (age 24.38 ± 3.84 years, height 1.65 ± 0.05 m, weight 66.98 ± 7.88 kg) and twelve males (age 25.33 ± 4.72 years, height 1.76 ± 0.07 m, weight 76.97 ± 9.15 kg). All participants were deemed able to participate in physical activity via the PAR-Q+, were not allergic to adhesive tape, and had not fallen during tasks of daily living in the 12 months prior to their data collection session. All participants were able to complete the CTSIB-M with only 1 trial per condition.

Increasing Lean Magnitude of Successful Trials

Paired samples T-tests were used to determine how the compensatory stepping response changed as the perturbation magnitude increased by determining the difference between participants' initial lean angle and maximal lean angle. Pearson correlations were also used to better understand the relationship between variables in the restabilisation phase. The initial lean angle for all participants was 10%BW (N=25), while the maximal lean angles included 25%BW (N=8), 30%BW (N=8), 35%BW (N=8), and 40%BW (N=1). The means and standard deviations for all measures at the initial and maximal lean angles can be found in Table 3.1.

Paired Samples T-Tests

The paired samples T-tests were run for all variables found in Table 3.1 comparing participants' initial lean angle and their maximal lean angle. From the initial lean angle to the maximal lean angle, there were significant increases in the AP variables of step length [$t(24) = -8.82$, $p < .001$], AP θ_d at P1 [$t(24) = -2.71$, $p = .012$], AP GRF at P1 [$t(24) = -2.98$, $p = .007$], AP θ_d at P2 [$t(24) = -3.18$, $p = .004$], and AP GRF at P2 [$t(24) = -3.72$, $p = .001$] while there were significant decreases for the AP MoS at perturbation onset [$t(24) = 25.28$, $p < .001$], AP MoS at toe-off [$t(24) = 23.08$, $p < .001$], AP MoS at foot-contact [$t(24) = 15.55$, $p < .001$], AP θ_d at foot-contact [$t(24) = 3.82$, $p = .001$], AP GRF at foot-contact [$t(24) = 15.22$, $p < .001$], AP MoS at P1 [$t(24) = 13.68$, $p < .001$], and AP MoS at P2 [$t(24) = 3.17$, $p = .004$]. There were no significant differences found for the AP measures of the time to AP P1 [$t(24) = -0.88$, $p = .385$] and the time to AP P2 [$t(24) = 0.01$, $p = .994$].

From the initial lean angle to the maximal lean angle, there were significant increases in the ML measures of step width [$t(24) = -2.84$, $p = .009$], ML θ_d at foot-contact [$t(24) = -2.73$, $p = .012$], ML θ_d at P1 [$t(24) = -4.31$, $p < .001$], and ML θ_d at P2 [$t(24) = -8.08$, $p < .001$] while there were significant decreases for the time to onset of ML asymmetry [$t(24) = 11.40$, $p < .001$], ML MoS at toe-off [$t(24) = 5.11$, $p < .001$], ML MoS at foot-contact [$t(24) = 7.58$, $p < .001$], ML GRF at foot-contact [$t(24) = 15.07$, $p < .001$], ML MoS at P1 [$t(24) = 5.77$, $p < .001$], time to P1 ML [$t(24) = 6.19$, $p < .001$], and ML MoS at P2 [$t(24) = 2.36$, $p = .026$]. There were no significant differences found for the ML measures of the ML MoS at perturbation onset [$t(24) = 0.86$, $p = .397$], ML GRF at P1 [$t(24) = 1.17$, $p = .252$], ML GRF at P2 [$t(24) = 1.20$, $p = .241$], and the time to P2 ML [$t(24) = -0.44$, $p = .667$].

Of the dependent variables not specific to a single direction, from the initial lean angle to the maximal lean angle, there was a significant increase in the time to restabilisation [$t(24) = -5.05$, $p < .001$] and significant decreases in the time to step onset [$t(24) = 8.06$, $p < .001$] and swing time [$t(24) = 5.46$, $p < .001$].

Table 3.1

Means and Standard Deviations of Variables in Participants' Initial and Maximal Lean Angles

Variables of Initial Step	Initial Lean Angle	Maximal Lean Angle
AP MoS at Perturbation Onset (m)***	-0.013 ± 0.039	-0.291 ± 0.056
ML MoS at Perturbation Onset (m)	0.190 ± 0.029	0.185 ± 0.036
Time to ML Asymmetry Onset (s)***	0.116 ± 0.035	0.010 ± 0.033
Time to Step Onset (s)***	0.383 ± 0.066	0.286 ± 0.023
AP MoS at Toe-Off (m)***	-0.154 ± 0.058	-0.565 ± 0.091
ML MoS at Toe-Off (m)***	0.203 ± 0.038	0.161 ± 0.038
Step Length (%BH)***	60.06 ± 8.517	79.52 ± 9.257
Step Width (%BH)**	29.99 ± 2.952	31.80 ± 4.304
Swing Time (s)***	0.179 ± 0.024	0.144 ± 0.024
Variables at Foot-Contact	**Initial Lean Angle**	**Maximal Lean Angle**
AP MoS at Foot-Contact (m)***	0.169 ± 0.044	-0.088 ± 0.072
ML MoS at Foot-Contact (m)***	0.115 ± 0.019	0.075 ± 0.026
AP θ_d at Foot-Contact (deg)**	-5.004 ± 2.723	-11.62 ± 8.443
ML θ_d at Foot-Contact (deg)**	-2.056 ± 1.146	1.066 ± 5.323
AP GRF at Foot-Contact (N)***	520.35 ± 94.25	173.72 ± 77.57
ML GRF at Foot-Contact (N)***	516.56 ± 92.41	167.78 ± 80.79
Variables at P1	**Initial Lean Angle**	**Maximal Lean Angle**
AP MoS at P1 (m)***	0.181 ± 0.043	0.018 ± 0.050
ML MoS at P1 (m)***	0.101 ± 0.019	0.071 ± 0.025
AP θ_d at P1 (deg)**	1.652 ± 1.015	3.581 ± 3.209
ML θ_d at P1 (deg)***	2.866 ± 0.859	5.010 ± 2.515
AP GRF at P1 (N)**	1015 ± 153.9	1178 ± 336.9
ML GRF at P1 (N)	303.1 ± 465.4	251.8 ± 453.2
Time to AP P1 (s)	0.048 ± 0.009	0.052 ± 0.019
Time to ML P1 (s)***	0.072 ± 0.020	0.038 ± 0.019
Variables at P2	**Initial Lean Angle**	**Maximal Lean Angle**
AP MoS at P2 (m)**	0.250 ± 0.050	0.204 ± 0.054
ML MoS at P2 (m)*	0.121 ± 0.019	0.106 ± 0.024
AP θ_d at P2 (deg)**	2.039 ± 0.879	3.218 ± 1.616
ML θ_d at P2 (deg)***	1.134 ± 0.719	2.538 ± 0.877
AP GRF at P2 (N)**	706.6 ± 96.42	771.9 ± 112.2
ML GRF at P2 (N)	733.5 ± 105.9	716.4 ± 111.5

Time to AP P2 (s)	0.318 ± 0.079	0.318 ± 0.114
Time to ML P2 (s)	0.361 ± 0.129	0.375 ± 0.103
Time to Restabilisation (s)***	3.464 ± 1.601	5.298 ± 2.054

Note. * $p < .05$, ** $p < .01$, *** $p < .001$

Failed Trials Compared to Successful Trials

Independent samples T-tests and binary logistic regressions were performed to understand if there were significant predictors of a successful or failed balance recovery response at a lean angle of 30%BW and 35%BW; these two lean angles were chosen as they, compared to lean angles of 40%BW and higher, had the largest sample size of successful and failed balance responses that were closest to a 50:50 split in successful:failed responses (Table 3.2). All participants (N=25, 100%) participated in trials at a lean angle of 30% BW; seventeen of those participants were successful (N=17, 68%; they achieved 3 successful trials and ≤ 1 failed trial) and the remaining eight participants failed (N=8, 32%; they had 2 failed trials and ≤2 successful trials).

The seventeen (N=17) participants who were successful at the 30%BW lean angle continued to participate in trials at a lean angle of 35%BW; of these seventeen participants at 35%BW (N=17, 100%), nine participants were successful (N=9, 53%) while eight participants failed (N=8, 47%). The 40%BW condition was not assessed because of the nine participants that performed trials at this lean angle (N=9, 100%), eight participants failed (N=8, 89%) while only one was successful (N=1, 11%). The 45%BW condition was also not assessed as only one participant performed trials at this lean angle.

Table 3.2

Distribution of Participants that Succeeded and Failed at Each Lean Angle

Lean Angle (%BW)	Total Participants	Participant Outcome #		Participant Outcome %	
		Successful	Failed	Successful	Failed
10	25	25	0	100	0
15	25	25	0	100	0
20	25	25	0	100	0
25	25	25	0	100	0
30	25	17	8	68	32
35	17	9	8	53	47
40	9	1	8	11	89
45	1	0	1	0	100

Note. Only successful participants at a lean angle performed trials for the subsequent lean angle.

Independent Samples T-Tests

Independent samples T-tests were performed to determine the differences between the successful and failed balance recovery responses for the variables found in Table 3.3. The variables measured at events following foot-contact (i.e., P1, P2, and restabilisation) were not included in this analysis as most failed trials had their point of failure occur prior to these events and thus could not be calculated, however, there were some exceptions to this as explained further down (Table 3.4). For the 30%BW lean angle, comparing the failed trials to the successful trials, there were significant increases in the ML MoS at toe-off ($t(23) = -2.16$, $p = .042$) and significant decreases in step length [$t(23) = 2.72$, $p = .012$], AP MoS at foot-contact [$t(23) = 5.06$, $p < .001$], ML MoS at foot-contact [$t(23) = 2.75$, $p = .011$], AP θ_d at foot-contact [$t(23) = 3.21$, $p = .004$], and ML θ_d at foot-contact [$t(23) = 4.05$, $p = .001$] for the failed trials.

For the 30%BW lean angle, there were no significant differences found for the AP MoS at perturbation onset [$t(23) = 0.48$, $p = .638$], ML MoS at perturbation onset [$t(23) = -1.08$, $p = .291$], the time to step onset [$t(23) = -1.62$, $p = .119$], AP MoS at toe-off [$t(23) = 1.04$, $p = .311$], step width [$t(23) = 1.38$, $p = .182$], swing time [$t(23) = 0.03$, $p = .977$], AP GRF at foot-contact [$t(23) = -0.58$, $p = .570$], and ML GRF at foot-contact [$t(23) = -0.37$, $p = .717$] between the successful and failed trials. The AP and ML MoS at perturbation onset having no significant differences between groups indicated the successful and failed participants had the same starting conditions for the perturbation.

For the 35%BW lean angle trials, for the failed trials as compared to the successful trials, there were significant increases in the AP GRF at foot-contact [$t(8.23) = -3.00$, $p = .017$] and ML GRF at foot-contact [$t(8.42) = -2.82$, $p = .021$] and significant decreases in step length [$t(15) = 2.48$, $p = .026$] and AP MoS at foot-contact [$t(7.76) = 3.48$, $p = .009$].

For the 35%BW lean angle, there were no significant differences found for the AP MoS at perturbation onset [$t(15) = -1.56$, $p = .139$], ML MoS at perturbation onset [$t(10.86) = -1.23$, $p = .246$], time to step onset [$t(9.39) = -1.57$, $p = .148$], AP MoS at toe-off [$t(15) = -1.05$, $p = .311$], ML MoS at toe-off [$t(15) = -1.10$, $p = .289$], step width [$t(15) = -0.50$, $p = .621$], swing time [$t(15) = 0.90$, $p = .381$], ML MoS at foot-contact [$t(15) = -0.06$, $p = .957$], AP θ_d at foot-contact [$t(15) = -1.11$, $p = .285$], and ML θ_d at foot-contact [$t(15) = 0.40$, $p = .695$] between the successful and failed trials. The AP and ML MoS at perturbation onset having no significant differences between groups indicated the successful and failed participants had the same starting conditions for the perturbation.

Table 3.3

Means and Standard Deviations of Variables in Successful and Failed Trials at Lean Angles of 30%BW and 35%BW

Variable	Lean Angle 30%BW		Lean Angle 35%BW	
	Successful	Failed	Successful	Failed
AP MoS at Perturbation Onset (m)	-0.282 ± 0.030	-0.289 ± 0.038	-0.344 ± 0.023	-0.320 ± 0.038
ML MoS at Perturbation Onset (m)	0.186 ± 0.029	0.200 ± 0.033	0.170 ± 0.048	0.191 ± 0.020
Time to ML Asymmetry Onset (s)	0.012 ± 0.036	0.029 ± 0.045	0.003 ± 0.010	0.499 ± 1.411
Time to Step Onset (s)	0.285 ± 0.021	0.298 ± 0.015	0.269 ± 0.010	0.282 ± 0.023
AP MoS at Toe-Off (m)	-0.550 ± 0.084	-0.587 ± 0.085	-0.623 ± 0.067	-0.588 ± 0.070
ML MoS at Toe-Off (m)*	0.152 ± 0.030	0.179 ± 0.024	0.142 ± 0.049	0.163 ± 0.022
Step Length (%BH)*†	82.80 ± 13.33	68.23 ± 10.38	78.22 ± 8.272	65.79 ± 12.27
Step Width (%BH)	31.34 ± 4.533	28.85 ± 3.350	31.31 ± 5.069	32.39 ± 3.444
Swing Time (s)	0.147 ± 0.024	0.147 ± 0.018	0.130 ± 0.021	0.121 ± 0.023
AP MoS at Foot-Contact (m)***††	-0.049 ± 0.053	-0.190 ± 0.086	-0.136 ± 0.018	-0.226 ± 0.071
ML MoS at Foot-Contact (m)*	0.075 ± 0.026	0.043 ± 0.029	0.075 ± 0.031	0.076 ± 0.029
AP θ_d at Foot-Contact (deg)**	-3.731 ± 10.06	-15.66 ± 4.014	-15.47 ± 6.278	-11.24 ± 9.330
ML θ_d at Foot-Contact (deg)**	2.221 ± 4.185	-5.831 ± 5.545	0.945 ± 7.515	-0.276 ± 4.484
AP GRF at Foot-Contact (N)†	188.4 ± 77.39	207.9 ± 82.48	118.0 ± 33.87	237.2 ± 107.7
ML GRF at Foot-Contact (N)†	183.2 ± 74.36	195.3 ± 82.12	107.6 ± 37.65	224.2 ± 111.4

Note. For the Lean Angle 30%BW, the data is derived from 17 Successful participants (N=17) and 8 Failed participants (N=8). For the Lean Angle 35%BW, the data is derived from 9 Successful participants (N=9) and 8 Failed participants (N=8).
* $p < .05$, ** $p < .01$, *** $p < .001$ for the Lean Angle 30%BW
† $p < .05$, †† $p < .01$, ††† $p < .001$ for the Lean Angle 35%BW

Additional T-tests were performed for the restabilisation-phase variables that acted as exceptions (Table 3.4). These included, at the 30%BW lean angle, variables measured at the AP and ML P1 and the time to AP and ML P1 which were able to be calculated for seven (N=7) of the eight (N=8) participants who had a failed balance response. This also included, at the 35%BW lean angle, the ML P1 and time to ML P1 which were able to be calculated for all seventeen (N=17) participants whether they had successful or failed trials. At 30% BW lean

Table 3.4

Means and Standard Deviations of Exception Variables in Successful and Failed Trials at Lean Angles of 30%BW and 35%BW

Variable	Lean Angle 30%BW		Lean Angle 35%BW	
	Successful	Failed	Successful	Failed
AP MoS at P1 (m)	0.012 ± 0.041	-0.039 ± 0.110		
ML MoS at P1 (m)**	0.073 ± 0.025	0.023 ± 0.038	0.076 ± 0.022	0.078 ± 0.031
AP θ_d at P1 (deg)	5.645 ± 5.912	2.188 ± 9.055		
ML θ_d at P1 (deg)	5.131 ± 2.318	4.259 ± 0.906	5.990 ± 2.908	6.218 ± 1.849
AP GRF at P1 (N)*	946.1 ± 410.2	1221 ± 111.4		
ML GRF at P1 (N)	230.1 ± 464.5	282.6 ± 442.0	229.2 ± 440.2	175.5 ± 343.1
Time to P1 AP (s)*	0.037 ± 0.012	0.096 ± 0.046		
Time to P1 ML (s)*	0.037 ± 0.016	0.060 ± 0.025	0.043 ± 0.025	0.042 ± 0.014

Note. For the Lean Angle 30%BW, the successful trials included all seventeen (N=17) successful participants at that lean angle, and the failed trials included seven (N=7) of the eight (N=8) failed participants at that lean angle with the shown measures in their balance recovery response. For the Lean Angle 35%BW, data is derived from 9 successful participants (N=9) and 8 failed participants (N=8); P1 AP and Time to P1 AP were not present in enough of the failed participants to reasonably be compared to the successful participants.
* $p < .05$, ** $p < .01$, *** $p < .001$ for the Lean Angle 30%BW
† $p < .05$, †† $p < .01$, ††† $p < .001$ for the Lean Angle 35%BW

angle, for the failed trials as compared to the successful trials, there were significant increases for the time to P1 AP [$t(6.34) = -3.36$, $p = .014$], time to P1 ML [$t(22) = -2.72$, $p = .012$], and AP GRF at P1 [$t(20.49) = -2.55$, $p = .019$] and a significant decrease in the ML MoS at P1 [$t(22) = 3.86$, $p = .001$]. There were no significant differences found for the AP MoS at P1 [$t(6.70) = 1.17$, $p = .282$], AP θ_d at P1 [$t(22) = 1.11$, $p = .278$], ML θ_d at P1 [$t(20) = 1.44$, $p = .165$], and ML GRF at P1 [$t(22) = -0.26$, $p = .801$].

At the 35%BW lean angle, there were no significant differences found for any variable occurring at ML P1; ML MoS at P1 [$t(15) = -0.12$, $p = .907$], ML θ_d at P1 [$t(15) = -0.19$, $p = .852$], ML GRF at P1 [$t(15) = 0.28$, $p = .785$], and the time to ML P1 [$t(15) = 0.05$, $p = .961$].

Binary Logistic Regression of Anteroposterior Variables

For the binary logistic regression, the dependent variable is the balance response outcome (Successful = 0, Failed = 1). To fit with the assumption that independent variables must be linearly related to the dependent variable, only measures from the independent samples T-tests with a significance value of $p < .200$ will be considered as potential predictors of a failed balance recovery response. For the AP measures at a lean angle of 30%BW, this includes time to step onset, step length, AP MoS at foot-contact, and AP θ_d at foot-contact. For the 35%BW lean angle, this includes AP MoS at perturbation onset, time to step onset, step length, AP MoS at foot-contact, AP θ_d at foot-contact, and AP GRF at foot-contact.

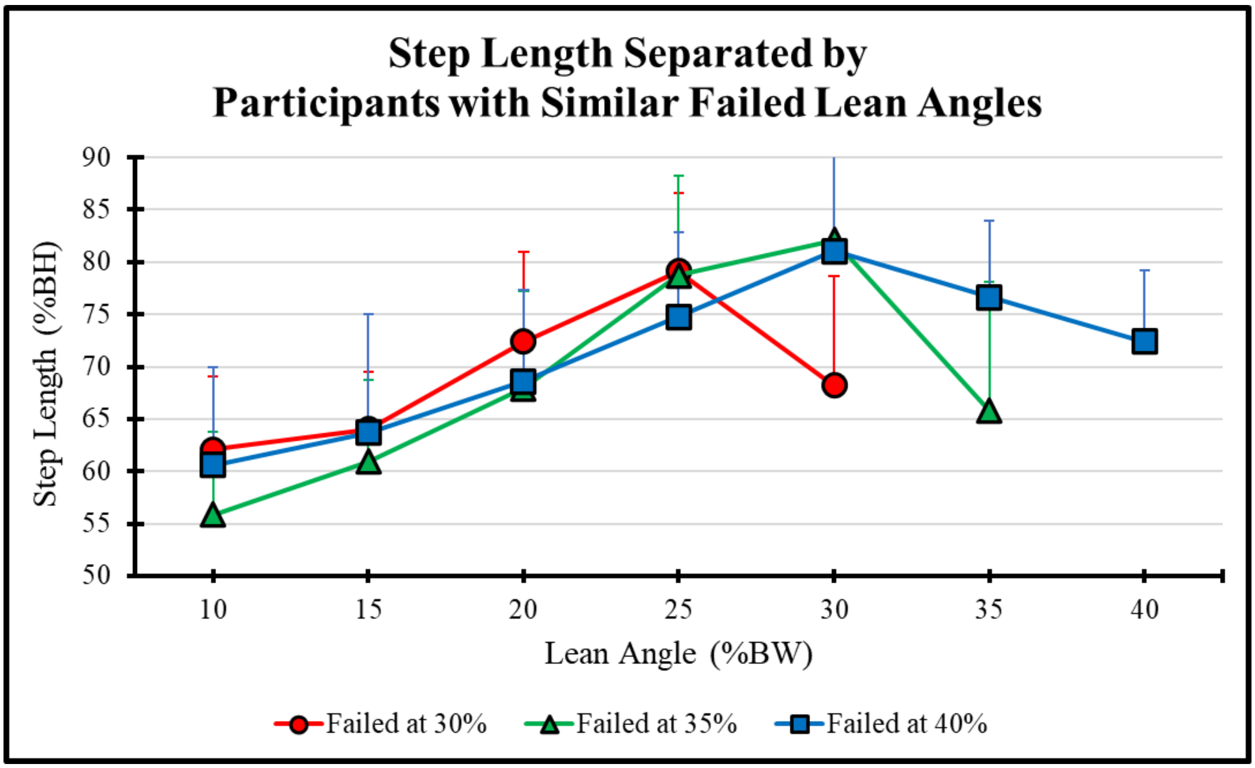
To check for the assumption of multicollinearity, Pearson correlations were performed on the variables of interest for the lean angles of 30%BW between successful and failed trials. There were significant positive correlations between step length and AP MoS at foot-contact ($r = .508$, $p = .010$), step length and AP θ_d at foot-contact ($r = .402$, $p = .046$), and AP MoS at foot-contact and AP θ_d at foot-contact ($r = .639$, $p = .001$); there were no significant correlations between time to step onset and all other measures including step length ($r = .360$, $p = .077$), AP MoS at foot-contact ($r = -.257$, $p = .215$), and AP θ_d at foot-contact ($r = -.117$, $p = .576$). Because the time to step onset was not significantly correlated with any other measure, it was included in the logistic regression model. Because the value of the AP MoS at foot-contact and AP θ_d at foot-contact are dependent on or correlated to the step length, step length was included in the logistic regression model while AP MoS at foot-contact and AP θ_d at foot-contact were excluded.

For the 35%BW lean angle between successful and failed trials, Pearson correlations revealed significant positive correlations between the AP MoS at perturbation onset and AP θ_d at foot-contact (r = .616, p = .009), time to step onset and AP GRF at foot-contact (r = .525, p = .030), and step length and AP MoS at foot-contact (r = .659, p = .004); all other correlations were not significant. The AP MoS at perturbation onset was excluded from the model as it was used to determine if the starting conditions between successful and failed groups were similar. Because the AP θ_d at foot-contact was not correlated to the other measures, it was included in the model. Because the value of the AP MoS at foot-contact is dependent on and correlated to the step length, step length was included in the logistic regression model while AP MoS at foot-contact was excluded. Because the effect of the AP GRF on the CoM kinematics is dependent on whether the AP θ_d is positive or negative and is dependent on that context for determining a success or fail, it was excluded from the model. The time to step onset was then included in the model as it was not correlated to any measures already included in the model.

The binary logistic regression for the 30%BW lean angle examined the significance of time to step onset and step length as predictors for a failed balance response task. The logistic regression model was statistically significant ($\chi^2(2) = 16.55$, $p < .001$). The model explained 67.8% (Nagelkerke R^2) of the variance in the balance response and correctly classified 92% of cases. A decrease in step length was more likely to result in a failed trial (OR = 0.82, 95%CI [0.69, 0.96]) and an increase in time to step onset was more likely to result in a failed trial (OR = 1.13, 95%CI [1.01, 1.27]). The binary logistic regression for the 35%BW lean angle examined the significance of time to step onset, step length , and AP θ_d at foot-contact as predictors for a failed balance response task. The logistic regression model was statistically significant ($\chi^2(3) = 12.64$, $p = .005$). The model explained 70.0% (Nagelkerke R^2) of the variance in the balance response and correctly classified 94.1% of cases. However, there was no significance in the odds of failing a balance response task due to individual changes in the time to step onset (OR = 1.162, 95%CI [0.96, 1.41]), step length (OR = 0.77, 95%CI [0.56, 1.06]), or AP θ_d at foot-contact (OR = 0.88, 95%CI [0.63, 1.25]).

Figure 3.1

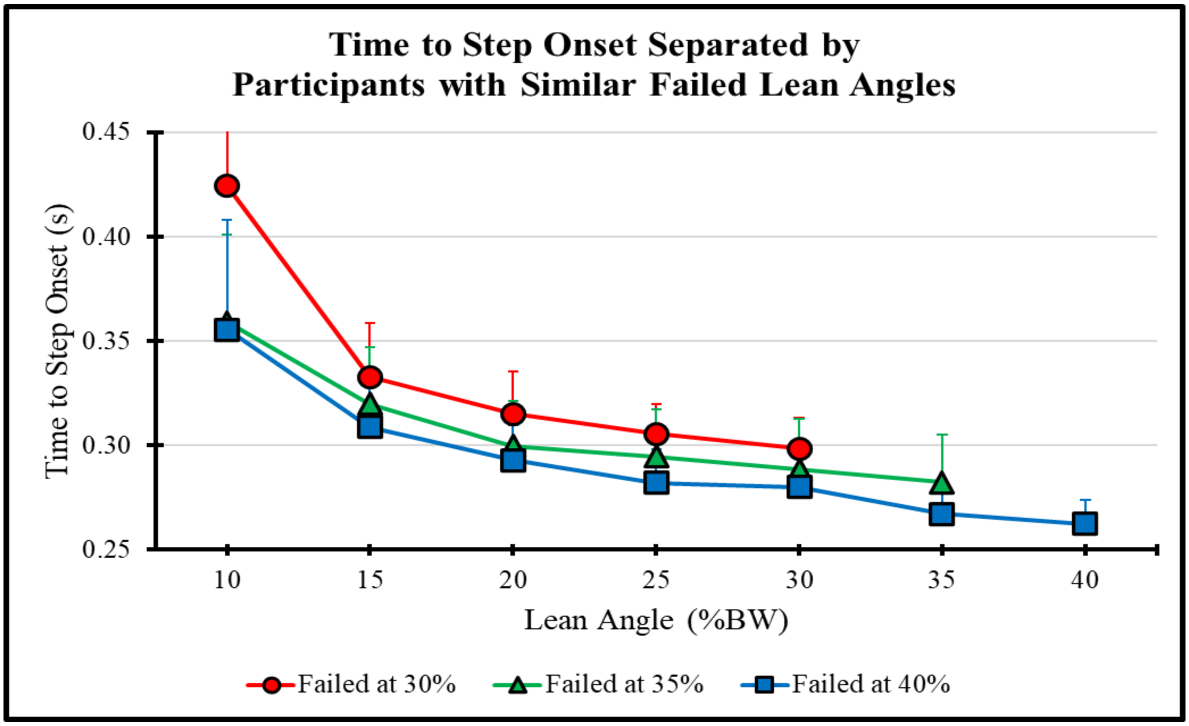
Changes in Step Length with Increasing Perturbation Magnitudes



Note. The "Failed at 30%" group contains data for the 8 (N=8) participants that achieved 2 failed trials at the 30%BW lean angle. Likewise, the "Failed at 35%" group contains data for 8 (N=8) participants, and the "Failed at 40%" group contains fata for another 8 (N=8) participants.

Figure 3.2

Changes in Time to Step Onset with Increasing Perturbation Magnitudes



Note. The “Failed at 30%” group contains data for the 8 (N=8) participants that achieved 2 failed trials at the 30%BW lean angle. Likewise, the “Failed at 35%” group contains data for 8 (N=8) participants, and the “Failed at 40%” group contains fata for another 8 (N=8) participants.

Binary Logistic Regression of Mediolateral Variables

To fit with the assumption that independent variables must be linearly related to the dependent variable, only measures from the independent samples T-tests with a significance value of $p < .200$ will be considered as potential predictors of a failed balance recovery response. For the ML measures at a lean angle of 30%BW, this includes time to step onset, ML MoS at toe-off, step width, ML MoS at foot-contact, and ML θ_d at foot-contact. For the 35%BW lean angle, this includes the time to step onset and ML GRF at foot-contact.

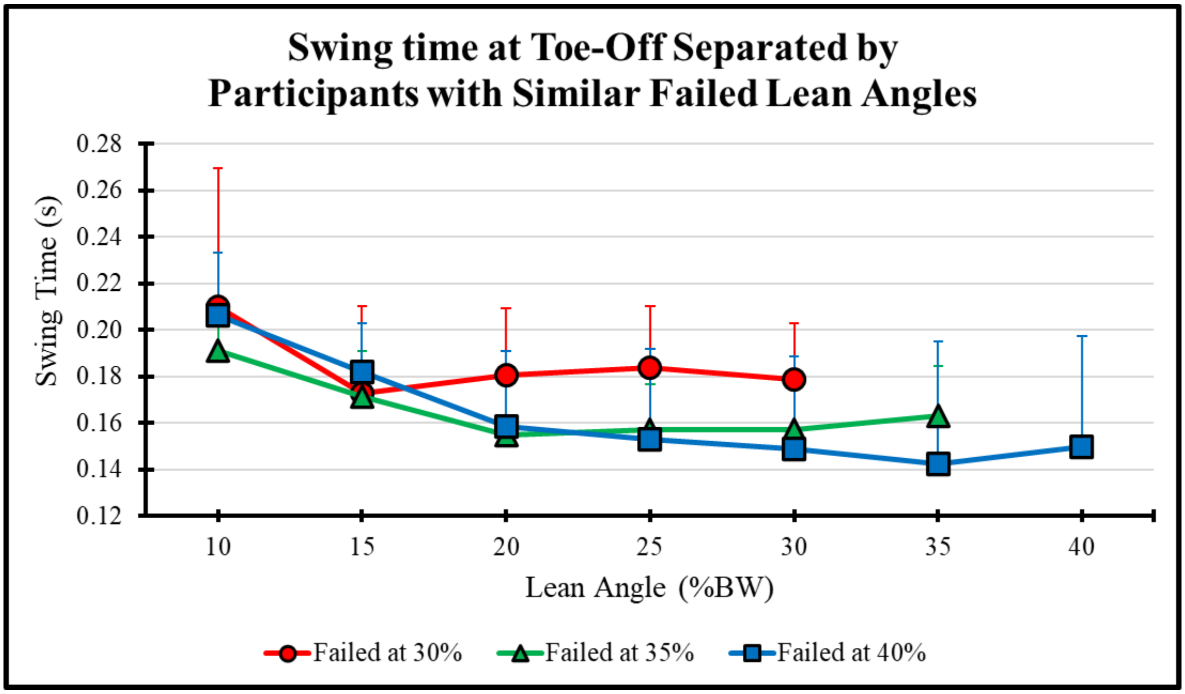
To check the assumptions of multicollinearity, Pearson correlations were performed on the variables of interest for the lean angles of 30%BW between successful and failed trials. There was a significant positive correlation between ML MoS at foot-contact and ML θ_d at foot-contact ($r = .752$, $p < .001$); there was a significant negative correlation between time to step onset and ML θ_d at foot-contact ($r = -.427$, $p = .033$); there were no other significant correlations found. The ML MoS at toe-off and step width were not significantly correlated with any other measure so they were included in the logistic regression model. Because the value of the ML MoS at foot-contact is dependent on the step width, it was excluded from the model. Lastly, because the ML θ_d at foot-contact revealed significant differences between successful and failed groups while the time to step onset did not, and because these measures are correlated, the ML θ_d at foot-contact was included in the model while the time to step onset was excluded.

For the 35%BW lean angle between successful and failed trials, Pearson correlations revealed a significant positive correlation between the time to step onset and ML GRF at foot-contact ($r = .490$, $p = .046$). Because the only 2 eligible variables were correlated to one another, they were excluded from the model. This meaning there was no binary logistic regression performed for the ML variables at the 35%BW lean angle.

The binary logistic regression for the 30%BW lean angle examined the significance of time to step onset, ML MoS at toe-off, and step width as predictors for a failed balance response task. The logistic regression model was statistically significant ($\chi^2(3) = 11.68$, $p = .009$). The model explained 52.2% (Nagelkerke R^2) of the variance in the balance response and correctly classified 80% of cases. An increase in the ML MoS at Toe-Off was more likely to result in a failed balance response task (OR = 1.94, 95%CI [1.02, 3.67]). However, there was no significance in the odds of failing a balance response task due to individual changes in the time to step onset (OR = 1.03, 95%CI [0.97, 1.09]) or the step width (OR = 0.68, 95%CI [0.39, 1.18]).

Figure 3.3

Changes in ML MoS at Toe-Off with Increasing Perturbation Magnitudes



Note. The "Failed at 30%" group contains data for the 8 (N=8) participants that achieved 2 failed trials at the 30%BW lean angle. Likewise, the "Failed at 35%" group contains data for 8 (N=8) participants, and the "Failed at 40%" group contains fata for another 8 (N=8) participants.

Chapter IV: Discussion

Purpose

The purpose of this study was to examine how the kinetics and kinematics of a balance response task change as a function of an increasing perturbation magnitude, as well as with a failed balance recovery response. The goals of the study were to identify the changes in the restabilisation-phase factors of a compensatory stepping response due to an increasing perturbation magnitude, as well as identifying what factors are likely to predict a failed balance recovery response. A maximum lean-and-release protocol was utilized as this involved inducing a continually increasing perturbation magnitude – in the form of an increasing lean angle – from which participants were required to recover their balance, until the point they could no longer recover their balance with a single step. To assess the changes as a result of an increasing forward lean angle, participants' initial and maximal lean angles – from which they could successfully recover balance– were compared within-subject. To assess the factors likely to predict a failed balance recovery response, the participants with failed trials were compared between-group to those with successful trials that occurred at the same lean angles.

Successful Balance Recovery Responses with Increasing Perturbation Magnitude

It was hypothesized that due to an increasing perturbation magnitude in successful balance recovery responses, participants would experience increasing instability at and following foot-contact which would necessitate a greater restabilisation response to recover from. It was expected that with an increased forward lean angle – which directly increases the forward CoM position and the gravitational moment about the CoM acting to increase its forward acceleration at perturbation onset – participants would have a larger AP destabilizing moment at foot-contact (via a larger, negative AP θ_d) and require a longer step length, larger AP θ_d and AP GRF at P1 and P2, and a shorter time to P1 and P2 in the AP direction; because the perturbation was increasing in the AP direction and there would be a reduced potential for producing an effective ML anticipatory postural adjustment (APA), it was expected that there would be an increased ML destabilizing moment at foot-contact (via larger, negative ML θ_d) but no differences in the step width, the ML θ_d and ML GRF at P1, and the time to P1 in the ML direction, though there would be an increased ML stabilizing moment at P2 (via larger ML θ_d and ML GRF) and a shorter time to P2 in the ML direction (as P2 may be modulated reactively in response to instability). The hypotheses that the AP stabilizing moments at P1 and P2 would increase (via larger AP θ_d and AP GRF) while only the ML stabilizing moment at P2 would increase (via larger ML θ_d and ML GRF) indicate a greater necessity to reorient the GRFnet in such a way to prioritize stabilizing the AP direction as opposed to, or prior to, the ML direction. The results supported some of these hypotheses and rejected others.

At the maximal lean angle, we observed a reduced AP θ_d (more negative) and a reduced AP GRF at foot-contact, a greater step length, and greater AP θ_d and AP GRF at P1 and P2, though no difference in the time to AP P1 and P2 were observed. With increased perturbation magnitudes, we observed an increased ML θ_d (from negative to positive) and a reduced ML GRF at foot-contact, a larger step width, an increased ML θ_d at P1 and P2, and a shorter time to ML P1, though no difference in the ML GRF at P1 and P2, and the time to ML P2.

Initial/Maximal Perturbations: From Perturbation Onset until Foot-Contact

As the lean angle increased, the initial conditions seen at perturbation onset showed the participants' AP MoS to decrease further in the negative direction as the CoM position was further anterior relative to the anterior border of the BoS, creating a larger AP moment arm between the CoP and CoM. Following the perturbation onset, this larger moment arm accelerated the CoM to a greater degree, increasing the CoM anterior position and velocity by foot-contact. To reduce the potential CoM position and velocity, participants produced a faster and larger initial step via a reduced time to step onset, reduced swing time, and longer step length. Despite this, in the maximal relative to initial trials, the AP MoS at foot-contact decreased and even became negative. It seems, with increasing perturbations, participants were unable to modulate their response in such a way to reduce or maintain the level of instability they face at foot-contact, but rather reduce the potential instability they will need to recover from (i.e., they reduce how much the instability increases by).

A negative value for the AP MoS at foot-contact in successful trials was not expected. Karaminidis, Arampatzis, and Mademli (2008) had similar trends of the AP MoS between small and large lean angles from perturbation onset to foot-contact, however they found the AP MoS at foot-contact in the large lean angles to remain a positive value. It should be noted their large lean angle was consistent for all participants as 33±3 %BW; it is possible they did not find a negative AP MoS at foot-contact as some participants might not have been at their maximal successful lean angle. The MoS measures the distance from the velocity-adjusted CoM (or X-CoM) to the borders of the BoS; even if the MoS is negative the CoM position (not adjusted for velocity) could still be within the borders of the BoS. At maximal lean angles that experience large accelerations of the CoM, it's possible the CoM velocity at foot-contact is so fast that the difference between the CoM position and the X-CoM is large enough that a person might have enough time to produce a sufficient stabilizing moment that decelerates the CoM to remain within the BoS.

The participants' ML MoS at perturbation onset was not different between lean angles (as the lean angle only increased in the anterior direction), yet their ML MoS at toe-off was still smaller at the larger lean angles. Despite participants producing a larger step width to reduce this instability, they still had a smaller (but positive) ML MoS at foot-contact. The earliest instance in the compensatory stepping response that did see a difference in the ML direction was the time to onset of ML asymmetry (time from perturbation onset until the vertical GRF differed between the left and right feet, which marks the beginning of the ML APA) which occurred significantly sooner in the maximal lean angle trials. The CoM experiencing larger accelerations in the maximal trials coupled with the reduced time to onset of ML asymmetry likely resulted in the CoM experiencing a larger, laterally directed acceleration sooner. Together, these results suggest that increasing AP perturbation magnitude does induce increased ML instability, however such instability may have its origins in the step initiation phase just following perturbation onset. This also suggests that an increased AP perturbation may increase the risk of a ML fall, which can have drastic implications for the older adult population. It has been found that older adults performing a compensatory step following a 10%BW lean angle perturbation experience ML instability during the restabilisation response and have a delayed reactive P2 response to reduce this instability (Singer et al., 2016). Forward and sideways falls are very commonly experienced by older adults (44% and 33% of all falls, respectively), with 47% of older adult falls resulting in impacting the hip/pelvis (Crenshaw et al., 2017). It is then especially worrying that increasing the anterior perturbation not only increases young adults' AP instability, but also their ML instability; with older adults already more unstable than young adults in the ML direction, and less able to recover from this instability at low perturbation magnitudes, it's likely this instability and poorer recovery response would be exacerbated with increased perturbation magnitudes, leading to a greater risk of AP and ML falls and injuries in the older adult population.

Diving deeper into the ML APA, it can also be observed – though it was not analysed statistically – that in the initial lean angle the sample's averaged ML MoS at toe-off (0.203 m ± 0.038) was slightly larger than the ML MoS at perturbation onset (0.190 m ± 0.029); in the maximal lean trials, the ML MoS at toe-off is smaller (0.161 m ± 0.038) than the ML MoS at perturbation onset (0.185 m ± 0.036), despite the shorter time to step onset. It is possible this observation could be due to the reduction, or decreased effectiveness, of an anticipatory postural adjustment (APA) from the initial to maximal trials. In support of this, the initial and maximal lean angle trials had similar times from the onset of ML asymmetry to the step onset (i.e., similar times in which the ML APA would be acting). Despite this, the maximal lean angle trials' APA was less able to control the CoM kinematics in such a way as to reduce the ML instability experienced at toe-off and, subsequently, foot-contact. Research has shown that the APA is present with voluntary stepping and well-practiced, or low-magnitude, perturbation-evoked stepping, and absent in large-magnitude perturbation-evoked stepping (McIlroy & Maki, 1999). Typically, the APA would not have a significant effect on the CoM kinematics during a rapid compensatory step, though it is possible the initial lean angle of 10%BW was a low enough magnitude that the ML CoM kinematics benefitted from an APA. Conversely, at larger magnitudes that would not benefit from an APA, it's possible that initiating an APA may delay the step onset or reduce the AP restabilising response, resulting in greater instability at and following foot-contact that increases the risk of falling. Whether the initial and maximal lean angles produced an APA would need to be confirmed in a follow-up analysis.

Initial/Maximal Perturbations: Stabilizing and Destabilizing Moments at Foot-Contact

At the moment of foot-contact in the initial lean angle, the participants produced AP and ML destabilizing moments about the CoM, as evident by the negative AP and ML θ_d at foot-contact. These results align with previous work that has found a compensatory step from a 10%BW lean angle results in the GRFnet eccentricity at the moment of foot-contact to further accelerate the CoM in the AP and ML directions towards the anterior and lateral borders of the CoM, creating – for a short time – even more instability immediately following foot-contact (Singer et al., 2019, 2016). In the maximal lean angle, the participants still produce an AP destabilizing moment at foot-contact with decreased (more negative) AP θ_d and magnitude of the AP GRF, as well as producing a ML stabilizing moment with an increased (and now positive) ML θ_d and a decreased ML GRF.

As a moment of force is the product of a force and its perpendicular distance from the point of rotation, an increase in the force and/or the distance can result in an increased moment. The θ_d in this case acts as a proxy measure for the perpendicular distance the GRF acts about the CoM; the larger the angle, the greater the distance, and vice versa. The orientation (θ_d) of the GRF vector in the maximal lean angle was more conducive to producing a larger destabilizing moment, though the AP GRF at foot-contact compensates for this as its reduced magnitude prevents the AP destabilizing moment to further increase. While an AP destabilizing moment is still produced, this strategy effectively reduces the additional anterior acceleration experienced by the (already faster and more anteriorly positioned) CoM at and following foot-contact; consequently, this also lowers the demand of P1 and P2 in reducing the remaining instability to achieve restabilisation had this strategy not been implemented.

The ML θ_d at foot-contact was negative in the initial lean angle but became positive in the maximal lean angle, while the ML GRF at foot-contact decreased from the initial to maximal lean angle. This indicates that the participants produced a ML destabilizing moment at foot-contact in the initial trials and produced a ML stabilizing moment at foot-contact in the maximal trials. This rejects the hypothesis that the increased lean magnitude would result in a greater ML destabilizing moment at foot-contact. These results can likely be attributed to the maximal trials' increased step width, and the more forward lean angle causing greater increases in AP than ML instability. This would more easily allow the ML GRF to act at a lateral distance about the CoM at foot-contact, producing a ML stabilizing moment that acts to redirect and accelerate the CoM towards the participants' midsagittal plane. This may also act as a strategy to limit the stabilising moments needed in the ML direction at P1 and P2 to achieve restabilisation, so restabilising in the AP direction can be prioritized.

Initial/Maximal Perturbations: Control of Restabilisation at P1

The timepoints of P1 and P2 were determined based on the first (P1) and second (P2) positive peaks of the GRF eccentricity (θ_d) following foot-contact. Following the increased instability at foot-contact in the maximal lean angles – evident by the reduced AP and ML MoS and increased AP destabilising moment at foot-contact – the participants required a more demanding restabilisation response to recover stability. From the initial to maximal lean angles, participants had reduced (though positive) AP and ML MoS at P1, indicating more instability within that first 100ms following foot-contact. From the initial to maximal lean angles, participants also increased their AP and ML θ_d at the respective AP and ML P1, as well as an increased AP GRF at P1 though there was no significant difference in the ML GRF at P1. In both directions, this creates larger stabilising moments about the CoM acting to slow and maintain the CoM within the BoS. The time to AP P1 between the initial and maximal lean angles showed no significant difference while there was a significantly shorter time to ML P1 in the maximal lean angle.

Interestingly, though not analysed statistically, the rate of development of the AP stabilizing moment was seen to be faster in the maximal lean angle trials due to no difference in the time to AP P1 between lean angles, but larger differences in the AP θ_d and AP GRF measures from foot-contact to P1 in the maximal lean angle. The maximal lean angle, as compared to the initial, had significantly smaller measures of the AP θ_d and AP GRF at foot-contact, though significantly larger measures of the AP θ_d and AP GRF at P1. When combined with no differences found in the time to AP P1, this indicates the participants, from foot-contact until P1, had a faster rate of generation of the AP stabilizing moments about the CoM via a combination of reorienting and increasing the magnitude of the AP GRF. The larger perturbation magnitudes result in the participants being in a less stable state at foot-contact and P1 due to a faster and more anteriorly positioned CoM, requiring this much larger and faster generated AP stabilizing moment at P1 to slow the CoM and prevent it from travelling beyond the anterior border of the BoS.

The ML stabilizing moments also see an increase in the rate of generation of the ML stabilizing moment from foot-contact until P1, but through different means. In the maximal lean angle, as compared to the initial, the ML θ_d at foot-contact increased (from negative to positive) though the ML GRF at foot-contact decreased, while the ML θ_d at P1 increased while the ML GRF at P1 showed no significant difference. When comparing the ML θ_d and ML GRF from foot-contact to P1 (Table 3.1), the ML θ_d does increase from foot-contact to P1 in both lean magnitudes, though the ML GRF from foot-contact to P1 decreases in the initial lean angle and increases in the maximal lean angle. This results in the ML stabilizing moments at foot-contact and P1 to have increased in the maximal lean angle, though the increase of the ML stabilizing moment from foot-contact to P1 is similar in both lean magnitudes. The key difference is that the maximal lean angle has a reduced time to ML P1, meaning the similar increase in the ML stabilizing moment occurs in less time and thus participants had an increased rate of generation of the ML stabilizing moments from foot-contact to P1.

Due to its rapidity following foot-contact, previous research has suggested the GRF eccentricity (θ_d) at P1 may be regulated by preparatory lower limb muscle activation and limb stiffness prior to foot-contact (Singer et al., 2016, 2019). Research has also shown increased muscular activity levels, prior to foot-contact, for the triceps surae, biceps femoris, rectus femoris, and tibialis anterior muscles when recovering from larger lean angle perturbations (Thelen et al., 2000). This increased preparatory muscle activation would increase limb stiffness on foot-contact, which can increase the vertical component of the GRFnet, resulting in a more vertical orientation of the GRFnet. This, coupled with a larger step length/width, would consequently result in a greater AP and ML eccentricity of the GRF (θ_d) relative to the CoM. While the moment of foot-contact itself showed smaller values of the AP θ_d and AP and ML GRF in the maximal lean angles, this likely relates to why there was an increased AP and ML θ_d at P1, as well as a greater rate of generation of the AP and ML stabilising moments from foot-contact to P1. Increasing the vertical component of the GRF would consequently reduce the horizontal components; as the lean magnitude primarily increased in the AP direction, it is understandable that the ML horizontal forces would have been further reduced compared to the AP horizontal forces. This may be why there was an increase in the AP GRF at P1, and no significant difference in the ML GRF at P1.

Initial/Maximal Perturbations: Control of Restabilisation at P2

The GRF eccentricity (θ_d) at P2 may be regulated by reactive control of the applied forces in response to instability, which may allow it to scale with the amount of instability present (Singer et al., 2016, 2019). In the maximal lean angle, participants experienced reduced stability at foot-contact and P1 which consequently required a greater restabilisation response at P2. Despite the increased step length and width and increased AP and ML stabilizing moments at P1, participants at their maximal lean angle – relative to their initial – experienced significantly smaller AP and ML MoS at the respective AP and ML P2. To reactively respond to the reduced stability, participants produced greater AP and ML stabilising moments at P2 via increased AP and ML θ_d at P2, and an increased AP GRF at P2. However, no differences were found for the ML GRF at P2 and the times to AP and ML P2. Despite the larger stabilising moments at P1 and P2, having to recover from more instability led to a significantly longer time to restabilisation.

Participants modulated their P1 and P2 response in the same way: the AP direction had a larger θ_d and GRF, while the ML direction had a larger θ_d but no difference in the GRF. The only difference being consistent timing of P2 for both directions when comparing initial to maximal lean angles. Although P1 and P2 have been suggested to be determined by proactive and reactive control, respectively, it’s possible the mechanisms of that control for P1 and P2 are similar, that being muscle activation. Research has found that the triceps surae, biceps femoris, rectus femoris, and tibialis anterior to be active just prior to foot-contact until about 500ms after foot-contact (Weaver, 2017). At both P1 and P2, the participants’ ability to modulate the orientation and magnitude of the GRF are likely controlled by the same muscles with similar mechanisms of control. Interestingly, previous research has found older adults, as compared to young adults, delay the deactivation of those same muscles following foot-contact (Thelen et al., 2000) and they have a longer time to ML P2 (Singer et al., 2016). It’s possible that the inability in producing an effective reactive stabilising response may be linked to lower limb muscle activation that controls the GRF orientation and magnitude.

Despite the increased rate of generation of the AP and ML stabilizing moments at P1, and the increased magnitude of the AP stabilizing moment at P1, participants were still required to generate larger AP and ML stabilizing moments about the CoM at P2 to achieve restabilisation. It is possible that the increased perturbation magnitude and subsequent increased velocity of the CoM was unable to be slowed enough by the stabilizing moments at P1 to allow for maintaining a similar response at P2 between lean magnitudes, requiring the P2 response at the maximal lean angle to increase accordingly. In this case, the orientation and magnitude of the GRF at P2 would be scaled reactively in response to the instability experienced through the restabilisaiton phase. It’s also possible that the P1 and P2 response scale together such that they can offset one another’s burdens of reducing instability; in other words, in achieving restabilisation, an increased P1 allows for a smaller reactive response at P2, while a larger P2 allows for a smaller proactive response at P1. By scaling together, the individual demands of P1 and P2 would be reduced.

These stabilizing moments occur with consistent timing between the two lean magnitudes. No differences were found for the time to AP and ML P2 between the initial and maximal lean angles. This suggests that regardless of the perturbation magnitude, the level of stability, or the previous restabilisation responses and their timing, the reactive response of P2 occurs with consistent timing in young adults. The CTSIB-M showed none of the healthy young adult participants to have issues with the various balance-related sensory systems, so it's possible the consistent timing of P2 (which may be regulated by reactive control in response to instability) corresponds to the participants ability to sense and react to feedback regarding the position and velocity of the CoM. This may explain why the response at P2 occurs with consistent timing between the initial and maximal lean magnitudes despite the significantly larger AP and ML stabilising moments produced at P2 in the maximal lean angle trials.

Although the AP and ML stabilizing moments at P2 were increased, there was still a significantly longer time to achieve restabilisation in the maximal lean angle trials. This is likely due to the significantly reduced AP and ML MoS at foot-contact, P1, and P2 and the larger responses required to slow the CoM to remain within the BoS. An increased time to restabilisation indicates that the CoM required a larger moment to act about it for a longer period of time, equating to an increased angular impulse about the CoM. However, if the moment acting to slow the CoM must be applied over too long a period of time, the CoM would eventually extend beyond the BoS due to the presence of an outward (away from the CoP) velocity of the CoM by the time the CoM reaches the borders of the BoS. At this point, the body would become unbalanced and a fall would be initiated.

Failed Balance Recovery Responses

It was expected that the balance response task would fail due to either the MoS being negative at the moment of foot-contact (as a result of too short a step length and/or too large of a CoM velocity) or the inability to produce sufficient stabilizing moments that act to slow the forward movement of the CoM (as a result of large destabilizing moments at foot-contact, decreased stabilizing moments at P1, and an increased time to, or absence of, P2). It was thus hypothesized that failed balance response tasks, compared to successful balance responses, would have a decreased step length, decreased AP and ML MoS at foot-contact, decreased AP and ML θ_d at foot-contact and P1, and a longer time to, or absence of, P2. It was further hypothesized that the predictors of a failed balance response task would be a decreased step length and AP θ_d at foot-contact. The first hypothesis was partially supported in that the failed balance responses, compared to the successful responses, had a decreased step length, a decreased (more negative) AP MoS at foot-contact, decreased AP θ_d at foot-contact (indicating a larger destabilising moment), as well as the absence of a P2 response. However, in the failed balance response trials that did have a P1 response (as not all did), there were no differences found for the AP or ML θ_d at the respective AP P1 and ML P1 between failed and successful trials. The second hypothesis was partially supported as a decreased step length was a predictor for a failed balance response task and a decreased AP θ_d at foot-contact contributed towards a significant model that predicted failed balance responses. It was also found that an increased time to step onset and an increased ML MoS at toe-off were significant predictors for a failed balance response task.

Failed Balance Recovery: From Perturbation Onset until Foot-Contact

The following results pertain to both the 30%BW and 35%BW lean angles. Comparing participants with failed trials to participants with successful trials at the same lean angle revealed no significant differences in the AP and ML MoS at perturbation onset, indicating that the initial conditions between those who succeeded and those who failed were the same. This further indicates that the reason a participant failed was due to the individual differences in their own balance response following the perturbation. No significant differences were found in the time to step onset and, subsequently, the AP MoS at toe-off; with a similar time from perturbation onset to toe-off, the CoM likely experienced similar accelerations such that the AP X-CoM was in a similar position relative to the anterior border of the BoS in both successful and failed trials.

Interestingly, at a lean angle of 30%BW, there was a significant difference in the ML MoS at toe-off between successful and failed groups, though it was the failed group that had the larger (more positive) value; a larger, positive value of the MoS indicates that the X-CoM is further inside the BoS and in a more stable state, indicating the failed group was in a more ML stable position at toe-off compared to the successful group. The 35%BW lean angle showed no significant difference of the ML MoS at toe-off between the successful and failed groups. It’s possible that, at the 30%BW lean angle, the failed group attempted to create a ML APA, resulting in their ML MoS at toe-off to be greater compared to the successful group. Contrary to this idea, the time to onset of ML asymmetry and time to step onset showed no significant differences at the 30%BW lean angle. Furthermore, though not analysed statistically, for both the successful and failed participants at the 30%BW lean angle, the ML MoS decreased from perturbation onset to toe-off. However, it’s also possible that the limited time to produce a ML APA at the 30%BW lean angle would not allow for maintaining or increasing the ML MoS from perturbation onset to toe-off, but rather limit its reduction. Whether this strategy was attempted by the failed group would need to be examined in a follow-up analysis.

Despite a similar AP MoS at toe-off, time to step onset, and swing time in the 30%BW and 35%BW lean angles, the failed group still had a shorter step length; this may be due to the failed group having a slower swing velocity, decreased flexibility, or reduced generation of propulsive forces of the stepping limb. With the shorter step length, the failed group also exhibited a decreased (or more negative) AP MoS at foot-contact. As the AP MoS at toe-off and swing time exhibited no differences, it's likely the AP X-CoM at foot-contact displaced similarly for both groups; a smaller step length with a similar position of the AP X-CoM would result in the decreased AP MoS seen in the failed group, placing the body in an imbalanced, unstable state at foot-contact. This supports the hypothesis that a fall is likely to occur due to a decreased step length resulting in a decreased (more negative) AP MoS value at foot-contact. This negative AP MoS at foot-contact was similarly seen in the successful maximal lean angle trials, though with a smaller value (less negative). These results, like the maximal lean angle, may suggest that it is still possible to recover balance from a negative MoS value without taking an extra step, though only if below some threshold.

At the 30%BW lean angle, the failed group also had a smaller, though still positive, ML MoS at foot-contact; there was no difference in this value for the 35%BW lean angle. As there were no significant differences in the time to step onset, swing time, or step width between groups, and there was a larger ML MoS at toe-off, it is unsure why the failed group exhibited a significantly smaller ML MoS at foot-contact at the 30%BW lean angle. One possible explanation could be due to the small, though non-significant, differences seen in step width. Compared to the successful group, the failed group had a non-significantly smaller step width at the 30%BW lean angle and a non-significantly larger step width at the 35%BW lean angle; this may explain why the 30%BW lean angle resulted in a difference in the ML MoS at foot-contact while the 35%BW lean angle did not. However, correlations between the step width and ML MoS at foot-contact for the 30%BW lean angle showed no significance.

At both the 30%BW and 35%BW lean angles, the failed group experienced larger AP and ML destabilizing moments at foot-contact compared to the successful group. At the 30%BW lean angle, this was due to the failed group having a decreased (more negative) AP θ_d at foot-contact and decreased (from positive to negative) ML θ_d at foot-contact, though no differences in the AP and ML GRF at foot-contact. At the 35%BW lean angle, this was due to the failed group having negative values for the AP and ML θ_d at foot-contact (though no significant differences) and increased AP and ML GRF at foot-contact. Either way the failed group experienced larger AP and ML destabilising moments at foot-contact, but through different mechanisms dependent on the lean angle; at 30% due to an inability to effectively reorient the GRF, and at 35% due to an inability to control the magnitude of the GRF. The larger destabilizing moments act to further accelerate the CoM towards the borders of the BoS, which is already reduced in the failed group via the smaller step length and reduced AP MoS at foot-contact, creating more instability in the already less stable participants. As compared to the successful group, this leaves the failed group in a significantly worse position to recover their balance and, if not already too unstable or imbalanced, would require a significantly larger restabilisation response at P1 and P2 to maintain or regain balance.

Failed Balance Recovery: Control of Stability During Restabilisation

Most failed trials had their point of failure (the first sample of data when the second stepping limb lifted off the ground or when the harness supported ≥ 20% of the participants' bodyweight) occur very quickly after foot-contact and, for many trials, prior to the events of P1 and P2. However, there were some exceptions to this. At the 30%BW lean angle, 7 of the 8 participants with failed responses contained data for AP and ML P1. At the 35%BW lean angle, all failed trials contained data for a ML P1.

At the 30%BW lean angle, participants with failed balance recovery responses, compared to those with successful trials, had an increased time to AP and ML P1, a decreased ML MoS at P1, and an increased AP GRF at P1, though there were no differences in the AP MoS at P1, AP and ML θ_d at P1, and the ML GRF at P1. Although there were no significant differences between successful and failed groups for AP MoS at P1, it should be noted that the successful group had a positive value for it while the failed group had a negative value. The results from the 30%BW lean angle indicate that, following the instability of a decreased AP MoS and larger destabilizing moments at foot-contact, the failed group did not struggle in generating a large enough stabilizing moment at P1, but rather the difficulty lied in generating that stabilizing moment quickly. This longer time may be due to the extra effort required from the failed group to reorient the GRF to move anterior and lateral to the CoM. Interestingly, though not analysed statistically, at the 30%BW lean angle, the successful response with the slowest time to AP P1 (0.055s) was still faster than the failed response with the fastest time to AP P1 (0.060s); this trend was not observed for the time to ML P1. Coupled with the larger AP destabilizing moment at foot-contact that acts to further accelerate the CoM, the extra time taken to generate AP and ML stabilizing moments is extra time the CoM has to increase its anterior and lateral position. To achieve P1, the failed group would have had to reorient the GRF to a larger degree to move the GRF anterior/lateral to the further positioned CoM. The decreased step length also hinders this endeavor as it limits the maximal potential anterior displacement of the CoP that the GRF may originate from. Despite this, the failed group may have generated a larger AP stabilizing moment at AP P1 compared to the successful group due to the larger AP GRF at P1. However, the larger stabilizing moment at P1 for the failed group was still not sufficient in maintaining balance, likely due to it having an insignificant effect on the CoM kinematics (the CoM was already positioned too far, and was moving too fast, anteriorly). The successful group would not have needed as large a stabilizing moment to maintain balance due to their increased step length, increased AP MoS at foot-contact, decreased AP destabilizing moment at foot-contact, and decreased time to AP P1.

It was also found that as the perturbation magnitude and instability in restabilisation increased, the capabilities of reactively responding to said instability were diminished. At the 30%BW lean angle, of the 8 participants with failed balance recovery responses, 5 participants produced an AP P2 and, of those 5, 3 participants produced an ML P2; though not analysed statistically, they clearly were not sufficient in preventing participants from failing to recover balance. At the larger perturbation magnitude of a 35%BW lean angle, none of the 8 participants with failed responses produced a P2 in either direction. This could even be due to the inability of P1 in slowing the CoM such that P2 could not generate a large or fast enough stabilizing moment to prevent the CoM from leaving the BoS; deficiencies in the ability to generate an effective stabilizing moment at P1 further reduces the effectiveness, and ability, in generating an effective stabilizing moment at P2. This both supports and contradicts previous findings indicating that effective proactive mechanisms at P1 may reduce the need for reactive mechanisms at P2, and that P2 may be capable of reducing the instability following an ineffective P1 (Rawal & Singer, 2021); this may be the case in lower perturbation magnitudes or during gait, but when restabilizing from a large perturbation magnitude, an ineffective P1 may result in the inability to produce a P2, and an effective P1 may or may not result in an effective P2. There is something to be said that, in situations comparing multiple successful stepping responses at large perturbation magnitudes, a larger P1 will result in a decreased P2 (as less reactive control is required to regain stability). However, this may also point to there being a certain perturbation threshold in which the ability to generate an effective reactive response of P2 is dependent on producing an effective P1. Overall, this indicates that the balance response task acts in a sequential, or successive, manner where one event can only occur if the events prior to it were effective. In other words, the restabilisation phase may only play a significant role in balance recovery when the initial properties of the balance response are sufficient, such as having a large step length or an increased stabilizing moment (or reduced destabilizing moment) at foot-contact that allow for generating a P1 and, subsequently, a P2.

Predictors of a Failed Balance Recovery Response

The logistic regression analysis was significant for the measures in the AP direction at both lean angles, though only significant at a 30%BW lean angle for the ML direction. The model looking at the 30%BW lean angle AP measures assessed the predictors of time to step onset and step length, both were significant in their ability to predict a failed balance response. With an increased time to step onset and a decreased step length, there is an increased likelihood of failing to recover balance following a forward perturbation. A reduced step length results in a shorter BoS, a decreased AP MoS at foot-contact, and potentially a decreased AP θ_d at foot-contact, which all act to put the body into a more unstable state. A longer time to step onset would provide the CoM more time to increase its forward velocity prior to foot-contact.

The model for the 35%BW lean angle AP measures assessed the predictors of time to step onset, step length, and AP θ_d at foot-contact, while the model was significant in its ability to predict a failed balance response, the individual variables were not significant. While the individual factor of AP θ_d at foot-contact was not significant in predicting a failed balance response task, the overall model was significant and able to predict 94.1% of cases, so it should be noted that a decreased AP θ_d at foot-contact did contribute to the model's ability to predict a failed balance response task. Decreases in the AP θ_d at foot-contact would result in a larger AP destabilizing moment at foot-contact that acts to further accelerate the CoM anteriorly, making it harder to maintain balance by slowing down a faster moving CoM.

At a lean angle of 30%BW, the logistic regression for the ML measures assessed the potential predictors of time to step onset, ML MoS at toe-off, and step width. The overall model and the ML MoS at toe-off were significant in their ability to predict a failed balance response, while the individual predictors of time to step onset and step width were not. With an increased ML MoS at toe-off, there is an increased likelihood of failing to recover balance following a forward perturbation. As previously mentioned, though unconfirmed, it is possible the failed group in the 30%BW lean angle had a larger ML MoS at toe-off compared to the successful group due to an attempt at initiating a ML APA. In this case, while a larger ML MoS at toe-off would indicate increased ML stability at toe-off, it is possible this took attention away from prioritizing stabilization in the AP direction, partially contributing towards the inability to restabilize. These observations were not seen in the 35%BW lean angle, though that may be due to the larger perturbation magnitude necessitating less time spent in double support to recover balance. Whether this strategy was attempted by the failed participants at the 30%BW lean angle would need to be confirmed in a follow-up analysis.

The results suggest that one of the key reasons for a failed balance response is simply not taking a large and fast enough step. A reduced time from perturbation magnitude until foot-contact (which includes the time to step onset) lessens the acceleration experienced by the CoM, decreasing the CoM velocity at foot-contact. An increased step length acts to enlarge the BoS and allows for generating larger stabilizing moments (via increased θ_d) about the CoM; this both acts to catch the CoM and decelerate it to a greater degree. If the time to step onset and step length are insufficient, the velocity of the CoM at foot-contact will increase and the ability to catch and slow the CoM is reduced – which combined results in an even smaller, or more negative, AP MoS at foot-contact – resulting in an increased likelihood of a failed balance response task. This is supported in the literature comparing older and younger adults' compensatory step at their respective maximum successful lean angle in which young adults are able to recover at a significantly larger forward lean angle than older adults, with the key reason why being the young adults' larger step length (Hsiao-Wecksler & Robinovitch, 2007). Further research compared older adults who were stable after a compensatory step (took one step to recover balance) to those who were unstable (took multiple steps to recover balance) (Arampatzis et al., 2008). They found the stable group to have a positive AP MoS at foot-contact while the unstable group had a negative value. They also found, in a different study, that young adults had a larger step length and, consequently, a larger AP MoS at foot-contact compared to older adults, irrespective of perturbation magnitude (Karaminidis et al., 2008). They concluded that the state of stability at foot-contact determines the ability to recover balance with a single step (Arampatzis et al., 2008). These studies support the notion that the initial response up to foot-contact is important in producing a response that effectively limits the instability experienced at and following foot-contact enough that stabilising moments can be generated to reduce the remaining instability and achieve restabilisation.

Assumptions

There are some assumptions that were made during this study. First, it was assumed that all participants' maximal lean angle trials resulted in an equally comparable balance recovery response. The maximal lean angles seen in this study include 25%BW, 30%BW, 35%BW, and 40%BW. When comparing participants' initial and maximal lean angle trials, all initial trials were at a lean angle of 10%BW while the maximal lean angle trials had that larger range. It's very likely that those able to recover from a larger maximal lean angle exhibited a different balance response pattern than those performing at a smaller maximal lean angle. Regardless, these trials involved the largest perturbation magnitudes that participants were able to successfully recover from, so in a sense we were comparing participants' easiest successful trials to their hardest successful trials. So, while the magnitude of the difference between initial and maximal trials may be different between participants, the within-subject design allowed for looking at the direction of difference (i.e., this score increased from initial to maximal for all participants, even if the magnitude of change was inconsistent).

Second, it was assumed that all failed trials were equally comparable. There were two reasons someone could fail a trial. Either they took a second step or they were supported by the harness. These groupings can be further broken down. Various observations in the failed trials due to taking a second step found participants took an additional forward or lateral step with their stance limb, an additional forward or lateral step with their stepping limb, because they fell forward or sideways, or because they entered a flight phase immediately following perturbation onset. It was observed that participants who failed due to being supported by the harness exhibited the same responses as noted above, in addition to some trials resulting in restabilisation with a single step but were significantly supported by the harness all the same. The outcome of a failed response was varied and, depending on the specifics of the failed response, the measures of said responses may have as much variability. Regardless of the outcome in a failed trial, the takeaway is that they were not able to recover their balance with a single step and without the harness supporting them. Because of the large variety in the outcomes of a failed response, if a commonality could be found between these responses, such as a significantly shorter step length that is also a significant predictor of a failed balance response, that may further highlight the importance of these factors in producing an effective compensatory stepping response.

Limitations

There were limitations to the current study. First, there was a potential impact of the harness on participants' natural responses. The harness was supported to a ceiling-mounted rail system via a rope of adjustable length. The length was adjusted such that the participants were just able to touch the ground with their fingertips with the knees in full extension. This length prevented the participants from impacting the ground with any part of their body except for their feet. However, at large lean angles of 40%BW or more, it's possible that participants with failed trials may have been able to successfully recover balance with a large and deep enough step. As the rope's length prevented participants from being close to impacting their knees on the ground, it is possible that the deep step required to recover balance was limited by the length of the harness rope. It is difficult to determine an appropriate solution to this problem as increasing the length of the rope may allow a participant to step deep enough to successfully recover balance, but it would also increase the chance of a participant impacting the ground when a fall occurs which is simply unacceptable.

A second limitation is not examining data collected at every lean angle. This study examined the change in the balance response task as a function of perturbation magnitude by comparing participants' initial lean angle to their maximal lean angle, as well as what occurs when a balance response task does not achieve restabilisation by comparing successful and failed trials that occur at the same lean angle. With the goal of assessing how the compensatory stepping response changes with an increasing lean angle, it would be beneficial to examine how the response changes with each increase in the lean angle. As the smallest maximal lean angle in the data was at 25%BW and the largest failed lean angle that was analysed was at 35%BW, the statistical analyses used in this study did not look at any successful trials at the 15%BW and 20%BW lean angles, an no failed trials at the 40%BW and 45%BW lean angles. With this in mind, the statistical analyses comparing successful to failed trials did not examine the failed trials of 9 participants. Collecting a larger data set may not necessarily fix this issue either as this still may result in a disproportionate amount of failed to successful trials that can't be reasonably compared without inflating the type I error. However, a more thorough statistical model may help to alleviate these issues, such as a hierarchical linear model to investigate the response changes using all lean angle trials.

Third, there were an unequal number of trials collected between the successful and failed balance responses that were then compared to one another. There were 3 successful trials collected at each successful lean angle, and only 2 failed trials collected at each failed lean angle. In maximum lean-and-release protocols, previous literature has demarcated 2 failed trials at the same lean angle to indicate someone is no longer able to successfully recover their balance and data collection ends, however those studies only analysed successful trials. Calculating the means from 2 failed trials, as opposed to 3, reduces the reliability of the results as the averaged values of the various measures may be further from the true values, leading to potential Type 1 and 2 errors when being directly compared to the means from 3 successful trials. A solution to this would be that when a participant achieves 2 failed trials at the same lean angle, they undergo an additional trial at the same lean angle to collect a 3rd failed trial. Alternatively, the protocol may be changed in that a lean angle is considered to be successful or failed when a participant first achieves 3 successful trials or 3 failed trials, respectively.

Lastly, the participants had varying degrees of physical activity experiences and fitness levels that may have allowed some participants to perform better than others. For example, there was only 1 female participant that had 3 successful trials (and less than 2 failed trials) at the 35%BW lean angle; all other female trials failed at the 35%BW lean angle or lower. This participant also has extensive gymnastic experience that may have given them that edge. Ultimately, this study examined how healthy young adults modulated their compensatory stepping response with increasing perturbation magnitudes until the point they fail to recover balance. Regardless of the participants' background, by statistically analysing only participants initial, maximal, and failed lean angles, we ensure we are only looking at participants easiest successful trials, hardest successful trials, and the trials that caused them to fail.

Conclusion

The goal of the compensatory stepping response is to recover and maintain balance following a perturbation, which is accomplished by catching the CoM within the newly formed BoS and maintaining it within those borders. As the perturbation magnitude increases, more and more instability is introduced, requiring participants to initiate their compensatory step more quickly, take a faster and longer step, and produce larger stabilizing moments during restabilisation. From small to large perturbation magnitudes, young adults modulate their initial step characteristics to effectively limit how much their instability increases by at and following foot-contact, consequently limiting the increased demands of the restabilisation phase in reducing the remaining instability. This is similarly seen within the restabilisation response as P1 and P2 appear to act inversely from one another where a more effective P1 allows for a reduced response at P2, and a reduced response at P1 requires a more effective P2. However, with increasing perturbation magnitudes, young adults take a longer time to achieve restabilisation, indicating that if the CoM is being slowed over too long a period of time, it will eventually extend beyond the BoS and cause a fall.

The compensatory step is not always effective in its goal of recovering and maintaining balance, and a fall will occur when the compensatory step fails to achieve restabilisation. Failed balance recovery responses, as compared to successful, have a reduced step length, decreased and negative AP MoS at foot-contact, and larger destabilizing moments at foot-contact. This effectively results in an unstable person at foot-contact becoming even more unstable. With an ineffective initial step, there is a reduced ability to generate a restabilisation response with many failed trials lacking P1 and P2 responses. Lastly, variables that can predict if a compensatory step will fail includes a longer time to step onset, a shorter step length, and a larger ML MoS at toe-off, further indicating the importance of the initial step characteristics. Ultimately, it appears the best way to prevent a fall from happening is by taking a fast and long step which aids in generating appropriate stabilizing moments to slow the CoM to regain and maintain balance and stability.

Implications and Significance of Research

This research study will contribute to the base of literature by providing insight into how young adults modulate their compensatory step and restabilisation phase to successfully recover balance when faced with increasing levels of instability, as well as the mechanisms that contribute to a failed balance recovery response. This study is contributing to the limited research that has investigated the period following foot-contact in a compensatory step by examining how this period (the restabilisation phase) is modulated with increasing perturbation magnitudes. There is even less research that has analysed trials with failed balance recovery responses; examining these trials is crucial to understanding the mechanisms that govern falls as well as confirming speculations regarding these mechanisms that were derived from successful balance recovery responses. Additionally, this study provides predictors that can be used to determine the likelihood of a failed balance recovery response, or rather, the essential elements of a compensatory step to avoid failing to recover balance.

The information learned from this study can be valuable to researchers, health practitioners, and for fall-prevention programs to have a better understanding of what can determine a successful vs. failed balance response, as well as how restabilisation is modulated in response to larger perturbations. When recovering from a perturbation, it is important to take a fast and large step, as well as control the orientation and magnitude of the GRF to produce stabilising moments. While taking a large, fast step may be easier to train, there are many ways we can manipulate the orientation of the GRF during gait and compensatory stepping. For example, orthotics and certain footwear can be designed to manipulate the GRF orientation to reduce instability at the ankle, knee, hip, and more superior joints. However, as P1 and P2 may be controlled through muscle activation, it's possible certain training may result in an increased biological ability to reorient the GRF during restabilisation to generate larger stabilising moments. By training or manipulating these factors a person may reduce their risk of falls, and understanding the predictors of a failed balance response may improve screening for fall risk.

Future Directions

The current study investigated the dynamics of restabilisation with increased perturbation magnitudes and in failed balance responses. However, this study only collected data from healthy young adults, with no demographics-based between-group comparisons. In future studies it would be beneficial to investigate how these results differ between males and females, younger and older adults, and other special populations with impaired gait and balance control. As this study only increased the perturbation magnitude in the anterior direction, it would also be beneficial to replicate this study with an increasing backwards or mediolateral perturbation, or with a different perturbation-evoking paradigm.

As mentioned in the limitations, a follow-up study using a more thorough statistical model, such as a hierarchical linear model, will allow for analyzing every lean angle trial in determining how the balance response is modulated with increasing perturbation magnitudes beyond only the initial and maximal. And having collected kinetic and kinematic data, follow-up analyses of this dataset can investigate different measures such as joint moments, joint velocities, and range of motion among others. Most of all, the hope is for future research to implement this knowledge into a fall prevention program to determine how populations at risk can learn to modulate their compensatory step and restabilisation phase and, hopefully, reduce their risk of future falls.

Overall, it would be beneficial for more researchers to examine and analyze trials in which participants fail to recover balance as we will gain the best insight into why a fall occurs by observing when a fall occurs.

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Appendix

Appendix I: Informed Consent Form

Faculty of Kinesiology
& Recreation Management

RESEARCH PARTICIPATION INFORMATION AND CONSENT FORM

Research Project Title: Examining the biomechanical mechanisms of restabilisation-phase stability control of compensatory stepping with increasing perturbation magnitudes and failed balance recovery

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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 are a leading cause of injury among all ages and poise a significant health concern for many special populations. These falls can occur while playing sports, participating in other forms of physical activity, or even during daily tasks of living. Falls may lead to a fracture, a ligament sprain, a muscle strain, etc. which can reduce independent mobility, increase future risk of falls, and can impact mental well being. Such reduction of individual independence can place a large burden on family members, caregivers, and the Canadian healthcare system.

When faced with a balance and/or stability issue a balance recovery response is often used to avoid falling and impacting the ground. This is commonly in the form of a compensatory step in which, as it sounds, a step is taken to recover one's balance and stability after being put off balance (or perturbed). However, after the foot contacts the ground (termed the restabilisation phase), there is still a chance for instability to arise and, subsequently, for a fall to occur. Despite considerable research, we lack specific understanding of how people lose their balance even after taking a compensatory step or, in other words, what happens when a compensatory step fails to restabilise. From a biomechanical perspective, the control of muscle forces from the lower limbs not only generates forward movement to start the compensatory step, but also controls the movement of the body's centre of mass when trying to restabilise. This control is challenging because the combined force output from the lower limbs must stop and then reverse the direction that the centre of mass is moving. As a balance recovery task becomes harder to perform (i.e., the force putting someone off balance increases), challenges in generating and/or controlling the lower limb force output may make it harder to slow and reverse the centre of mass movement, which can lead to instability.

The proposed work uses newly developed measures to better identify and understand the biomechanical mechanisms that contribute to a fall that occurs when a balance response task fails to recover balance. Identifying and understanding the specific mechanisms that contribute to instability and falls is important because it can improve the specificity of interventions that aim to reduce fall risk in special populations, such as targeted muscle strengthening or perturbation-based training programs where people can practice their compensatory stepping reaction.

The objectives of this study are to: (1) examine how the mechanisms of controlling stability during the restabilisation-phase change as the balance response task becomes more difficult (i.e., as the perturbation magnitude increases) and (2) identify the specific mechanisms that occur when a single compensatory step is not enough to restabilise.

Eligibility Requirements/Screening Procedures:

To be eligible for participating in this study, you will first be screened using a questionnaire regarding your general health. As this study seeks to recruit young healthy adults, it is necessary that those participants are well able to perform the tasks described below without any risk of injury. The Physical Activity Readiness Questionnaire (PAR-Q+) will ask 7 general health questions; this questionnaire helps determine if an individual is cleared to participate in physical activity. If you are not cleared to participate in physical activity, you will be excluded from participating in this study. In addition, you will be asked if you have a known allergy or sensitivity to adhesive tape/bandages; if yes, you will be excluded from participating in this study. Please understand the years of age of interest are 18-35 years.

Surveys and Questionnaires:

Prior to beginning the data collection session, you will be asked to complete a few questionnaires. These will include the Physical Activity and Sedentary Behavior Questionnaire (PASB-Q), a 12-month falls history questionnaire, the International Falls Efficacy Scale (FES-I), and a demographics questionnaire. The PASB-Q asks questions related to how much and what kind of physical activity you regularly participate in, as well as how much and what kind of sedentary behavior you regularly participate in. The 12-month falls history questionnaire, as the name suggests, asks questions regarding how often you have fallen in the past year, as well as how. The FES-I asks questions regarding how concerned your about falling across different tasks of daily living. The demographics questionnaire will gather info on your age, height, weight, sex assigned at birth, and gender. These questionnaires may be provided to you via email (from the principal investigator) prior to your data collection session. You may choose to fill these out by hand or on the computer and if so, you may either bring the physical copies to the data collection session or email the completed documents to the principal investigator. Filling these forms beforehand is simply to save time during the data collection session itself. Otherwise, physical copies will be available to fill out in the lab when you arrive for the data collection session.

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 body’s centre of mass and the force transmitted across your joints).

Reflective markers will be placed on both sides of your body ranging from the feet to hips, hands to shoulders, the sternum, and the ears. The reflective markers will be placed on top of your clothing using double-sided tape. There will also be rigid plastic pieces which contain clusters of four markers that will be placed, via a Velcro strap, on the shins, thighs, lower back, upper back, upper arms, and the forearms. Lastly, there will be a Velcro head strap which contains 8 markers that will be worn like a headband. If you would like to see a visualization of the marker placement on a person, please contact the Principal Investigator.

After having the markers placed, 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.

For all of the trials you choose to participate in that are outlined below, you will be asked to wear a harness that will be securely attached, via a rope, to an overhead support rail that is rigidly mounted in the laboratory ceiling; this overhead connection will prevent you from falling and impacting the ground. For extra precaution, crash mats will be placed around you during these trials. The following data collection trials will then be conducted:

1. Stationary, Quiet Standing:

In this condition, you will be standing upright with your feet side-by-side and arms hanging by your sides. Each foot will be placed on a separate force platform (a tool used to measure the forces you exert into the ground). You will be asked to stand stationary for a period of 60 seconds. 2 trials of stationary standing with your feet side-by-side will be collected. You will then be asked to do 2 more trials, though now standing with your feet in a staggered position (one foot in front of the other), with each foot placed on a separate force platform. Each trial will have a different foot placed in the forward position. If you are unable to remain motionless, you will be asked to move back into the "quiet standing" position until the end of the trial. You will be given a break (as much time as you need) after each trial.

2. Modified Clinical Test of Sensory Interaction in Balance (CTSIB-M):

In this condition, you will be asked to stand still for 30 seconds across 4 different conditions: (1) eyes open, standing on a firm surface (the force platforms); (2) eyes closed, firm surface; (3) eyes open, standing on a foam surface; (4) eyes closed, foam surface. The trial will end when you (a) open your eyes during an ‘eyes closed’ condition, (b) raise your arms from your sides, (c) lose balance and require assistance from the harness, (d) successfully stand still for 30 seconds, or (e) verbally indicate that you would like to stop. You will be given up to 3 trials per condition. If you are able to stand still for 30 seconds, then no other trials will be performed for that condition.

3. Perturbation-Evoked Stepping Conditions:

You will also be asked to take part in trials in which your standing balance will be perturbed (put off balance) by means of a custom built perturbation apparatus. In addition to the overhead support rail, the back of your harness will be attached to a rigid steel frame via a metal cable. For the trials, you will be asked to stand with your feet side-by-side and to lean forward until the metal cable supports approximately 10% of your body weight. We will then release the metal cable to cause a forward balance perturbation that will be sufficient enough to cause a single forward step reaction. You will be asked to regain your balance using a single compensatory step and to maintain this new stance position for approximately 10 seconds. After three trials in which you successfully regain your balance with a single step, you will be asked to perform an additional set of 3 successful trials though with an increased lean angle of an additional 5% bodyweight being supported by the cable. In other words, the 2^{nd} set of trials will have you lean forward with 15% of your bodyweight being supported, the 3^{rd} set of trials will support 20% of your bodyweight, and so on. This process will repeat until you have 2 failed trials at the same lean angle where you are either unable to recover your balance with a single step or if the rope supports 20% of your bodyweight. At that point, the experimental trials will end. To reiterate, the harness being attached to the overhead support rail will prevent you from falling and impacting the ground and, in addition, we will have crash mats placed around you as an extra precaution. To discourage anticipation of the cable release, the cable will be released at random intervals after you lean forwards. In addition, 25% of experimental trials will consist of 'catch trials', in which the cable will not be released and you will not experience a perturbation; these will be randomly presented throughout the course of testing and will not count towards your successful or failed trials.

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; quiet standing trials takes approximately 10 minutes; the CTSIB-M trials takes approximately 5 minutes; perturbation-evoked stepping trials takes up to 60 minutes. You have the choice to end your participation at any point in the data collection session. You may refer to the "Changing Your Mind About Participation" section below for a better understanding of how to withdraw from the study.

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 a force platform. 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 and after being perturbed, there are no direct benefits to you from participating. However, completing the CTSIB-M could possibly direct your attention towards a specific sensory system that might cause you to have some balance difficulties, potentially preventing falls from occurring in the future.

Risks to Participation and Associated Safeguards

The quiet standing and CTSIB-M trials are safe to perform for special populations (such as older adults) even without a harness, and the perturbation-evoked stepping trials, when carried out correctly, pose no major safety risks. Nevertheless, during these trials you will be wearing a harness preventing an actual fall from occurring and thus avoiding impacting the ground. And as an additional precaution, crash mats will be placed around you during these trials.

As the perturbation-evoked stepping trials continue until you are unable to recover balance with just a single step, you are likely to experience a falling sensation. However, you will be wearing a harness preventing an actual fall from occurring, as well as crash mats placed around you for additional safety.

The perturbation-evoked stepping trials involves administering perturbations to bring you into a state of imbalance so as to elicit a balance recovery response (compensatory step). In addition, these perturbations are to increase in magnitude (lean angle increases) until you are unable to recover your balance with just a single step. As well as wearing a safety harness that's connected to an overhead support rail that acts to prevent a fall from actually occurring, to mitigate any risk, all components of the test will be explained and demonstrated to you by the principal investigator and the research assistant. You will attempt the task with the principal investigator and research assistant. In the case that your attempt indicated a misunderstanding of the instructions, another demonstration will be given, and you will be allowed a second attempt. Again, in addition to the safety harness, crash mats will be placed around you.

If you ask for your CTSIB-M results upon completion of the test, the researchers will share these with you. These results, if poor, may cause you to become fearful of losing your balance in everyday situations. This fear may result in minor psychological distress. If these results are requested, researchers will encourage you to not feel discouraged and will discuss with you the fact that these results are solely for research purposes. If you would like a clinical confirmation of your scores, a clinician such as a physical therapist should be sought out. A clinician, at this point, would be capable of helping you work to improve your balance, if you are interested in doing so.

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 the investigators in the room aware of this and testing will stop immediately.

Anonymity and Confidentiality of Data:

Each participant will be given a unique alphanumerical code to prevent your data from directly or indirectly identifying you. Using a participant ID key, your code will be linked to your name and contact information. The participant ID key and the bottom portion of the participant feedback form, which will contain your name, will be kept in a locked filing cabinet in the Research Supervisor's office for 5 years after the completion of the study (the Research Supervisor's office is located behind two locked doors). Only the Principal Investigator and the Research Supervisor will know and have access to where these forms are. Upon completion of the study, which is expected to occur December 2022, the participant ID key and the participant feedback form will be destroyed via a file shredder. This consent form and the PAR-Q+, which will contain your name and signature, will be kept in the same locked filing cabinet in the Research Supervisor's office for 5 years after the completion of the study. Only the Principal Investigator and the Research Supervisor will know and have access to where these forms are. 5 years following the completion of this study, this consent form and the PAR-Q+ will be destroyed via a file shredder. This is expected to occur December 2027.

The PASB-Q, Fall History Questionnaire & FES-I, and demographics questionnaire will only contain your alphanumerical code, which has no personally identifiable information. This data will be stored in a locked filing cabinet in the Research Supervisor's office for 5 years after the completion of the study, at which point the data will be destroyed via a file shredder; this is expected to occur December 2027.

The biomechanical data (motion analysis and force platform data) will include the initial calibration trial, the quiet standing trials, the CTSIB-M trials, and the perturbation-evoked stepping trials. As the motion analysis cameras only record the position of the reflective markers located on your body, this recording contains no identifiable data, though the file name itself will contain your alphanumerical code. There is also no way to identify you from the forces you apply to the force platforms, though the file name will contain your alphanumerical code. You will be identified only by the participant alphanumerical code, which contains no personally identifiable information. These codes contain only a number/letter and cannot be linked back to any specific person without the participant ID key.

The Principal Investigator, Research Supervisor, and Research Assistant will have access to the biomechanical 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's information. The biomechanical data, which are coded via their file name, will be retained indefinitely on a password-protected computer in the Research Supervisor'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 when you participate in the study. This will be in the form of 2 Winnipeg transit bus tickets.

Changing Your Mind About Participation

You may withdraw from this study at any time without any negative consequences. You may withdraw before the study via email, phone, or in-person. To do so, indicate this to the private investigator and/or research supervisor stating, 'I no longer wish to participate in this study', or a similar statement. You may also communicate this during the data collection session at any point to anyone on the research team by stating, '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 easiest way to locate your data and delete it. Please note that there will be an inability to withdraw any data collection by December 2022. (12/22). At this point in time, all data collected will be used for discussion and will be analyzed to view level of significance. The data will then be published and at that time will be public data. However, no personal information will be included to identify any of the participants.

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 takes 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 April 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 or the journal Gait and Posture). 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. This study will also be used to form a master's thesis manuscript which will be published and publicly available on MSpace.

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 Board at the University of Manitoba, Fort Garry campus. If you have any concerns or complaints about this project you may contact any of the above-named 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 II: Physical Activity Readiness Questionnaire for Everyone (PAR-Q+)

2021 PAR-Q+

The Physical Activity Readiness Questionnaire for Everyone

The health benefits of regular physical activity are clear; more people should engage in physical activity every day of the week. Participating in physical activity is very safe for MOST people. This questionnaire will tell you whether it is necessary for you to seek further advice from your doctor OR a qualified exercise professional before becoming more physically active.

GENERAL HEALTH QUESTIONS

Please read the 7 questions below carefully and answer each one honestly: check YES or NO.	**YES**	**NO**
1) Has your doctor ever said that you have a heart condition ☐ **OR** high blood pressure ☐?	☐	☐
2) Do you feel pain in your chest at rest, during your daily activities of living, **OR** when you do physical activity?	☐	☐
3) Do you lose balance because of dizziness **OR** have you lost consciousness in the last 12 months? Please answer **NO** if your dizziness was associated with over-breathing (including during vigorous exercise).	☐	☐
4) Have you ever been diagnosed with another chronic medical condition (other than heart disease or high blood pressure)? **PLEASE LIST CONDITION(S) HERE:** ________	☐	☐
5) Are you currently taking prescribed medications for a chronic medical condition? **PLEASE LIST CONDITION(S) AND MEDICATIONS HERE:** ________	☐	☐
6) Do you currently have (or have had within the past 12 months) a bone, joint, or soft tissue (muscle, ligament, or tendon) problem that could be made worse by becoming more physically active? Please answer **NO** if you had a problem in the past, but it **does not limit your current ability** to be physically active. **PLEASE LIST CONDITION(S) HERE:** ________	☐	☐
7) Has your doctor ever said that you should only do medically supervised physical activity?	☐	☐

If you answered NO to all of the questions above, you are cleared for physical activity.
Please sign the PARTICIPANT DECLARATION. You do not need to complete Pages 2 and 3.

- Start becoming much more physically active – start slowly and build up gradually.
- Follow Global Physical Activity Guidelines for your age (https://www.who.int/publications/i/item/9789240015128).
- You may take part in a health and fitness appraisal.
- If you are over the age of 45 yr and NOT accustomed to regular vigorous to maximal effort exercise, consult a qualified exercise professional before engaging in this intensity of exercise.
- If you have any further questions, contact a qualified exercise professional.

PARTICIPANT DECLARATION
If you are less than the legal age required for consent or require the assent of a care provider, your parent, guardian or care provider must also sign this form.

I, the undersigned, have read, understood to my full satisfaction and completed this questionnaire. I acknowledge that this physical activity clearance is valid for a maximum of 12 months from the date it is completed and becomes invalid if my condition changes. I also acknowledge that the community/fitness center may retain a copy of this form for its records. In these instances, it will maintain the confidentiality of the same, complying with applicable law.

NAME ________________________ DATE ______________

SIGNATURE ________________________ WITNESS ______________

SIGNATURE OF PARENT/GUARDIAN/CARE PROVIDER ________________________

If you answered YES to one or more of the questions above, COMPLETE PAGES 2 AND 3.

Delay becoming more active if:

- You have a temporary illness such as a cold or fever; it is best to wait until you feel better.
- You are pregnant - talk to your health care practitioner, your physician, a qualified exercise professional, and/or complete the ePARmed-X+ at www.eparmedx.com before becoming more physically active.
- Your health changes - answer the questions on Pages 2 and 3 of this document and/or talk to your doctor or a qualified exercise professional before continuing with any physical activity program.

01-11-2020

2021 PAR-Q+

FOLLOW-UP QUESTIONS ABOUT YOUR MEDICAL CONDITION(S)

1. Do you have Arthritis, Osteoporosis, or Back Problems?

If the above condition(s) is/are present, answer questions 1a-1c — If **NO** ☐ go to question 2

1a.	Do you have difficulty controlling your condition with medications or other physician-prescribed therapies? (Answer **NO** if you are not currently taking medications or other treatments)	YES ☐ NO ☐
1b.	Do you have joint problems causing pain, a recent fracture or fracture caused by osteoporosis or cancer, displaced vertebra (e.g., spondylolisthesis), and/or spondylolysis/pars defect (a crack in the bony ring on the back of the spinal column)?	YES ☐ NO ☐
1c.	Have you had steroid injections or taken steroid tablets regularly for more than 3 months?	YES ☐ NO ☐

2. Do you currently have Cancer of any kind?

If the above condition(s) is/are present, answer questions 2a-2b — If **NO** ☐ go to question 3

2a.	Does your cancer diagnosis include any of the following types: lung/bronchogenic, multiple myeloma (cancer of plasma cells), head, and/or neck?	YES ☐ NO ☐
2b.	Are you currently receiving cancer therapy (such as chemotheraphy or radiotherapy)?	YES ☐ NO ☐

3. Do you have a Heart or Cardiovascular Condition? This includes Coronary Artery Disease, Heart Failure, Diagnosed Abnormality of Heart Rhythm

If the above condition(s) is/are present, answer questions 3a-3d — If **NO** ☐ go to question 4

3a.	Do you have difficulty controlling your condition with medications or other physician-prescribed therapies? (Answer **NO** if you are not currently taking medications or other treatments)	YES ☐ NO ☐
3b.	Do you have an irregular heart beat that requires medical management? (e.g., atrial fibrillation, premature ventricular contraction)	YES ☐ NO ☐
3c.	Do you have chronic heart failure?	YES ☐ NO ☐
3d.	Do you have diagnosed coronary artery (cardiovascular) disease and have not participated in regular physical activity in the last 2 months?	YES ☐ NO ☐

4. Do you currently have High Blood Pressure?

If the above condition(s) is/are present, answer questions 4a-4b — If **NO** ☐ go to question 5

4a.	Do you have difficulty controlling your condition with medications or other physician-prescribed therapies? (Answer **NO** if you are not currently taking medications or other treatments)	YES ☐ NO ☐
4b.	Do you have a resting blood pressure equal to or greater than 160/90 mmHg with or without medication? (Answer **YES** if you do not know your resting blood pressure)	YES ☐ NO ☐

5. Do you have any Metabolic Conditions? This includes Type 1 Diabetes, Type 2 Diabetes, Pre-Diabetes

If the above condition(s) is/are present, answer questions 5a-5e — If **NO** ☐ go to question 6

5a.	Do you often have difficulty controlling your blood sugar levels with foods, medications, or other physician-prescribed therapies?	YES ☐ NO ☐
5b.	Do you often suffer from signs and symptoms of low blood sugar (hypoglycemia) following exercise and/or during activities of daily living? Signs of hypoglycemia may include shakiness, nervousness, unusual irritability, abnormal sweating, dizziness or light-headedness, mental confusion, difficulty speaking, weakness, or sleepiness.	YES ☐ NO ☐
5c.	Do you have any signs or symptoms of diabetes complications such as heart or vascular disease and/or complications affecting your eyes, kidneys, **OR** the sensation in your toes and feet?	YES ☐ NO ☐
5d.	Do you have other metabolic conditions (such as current pregnancy-related diabetes, chronic kidney disease, or liver problems)?	YES ☐ NO ☐
5e.	Are you planning to engage in what for you is unusually high (or vigorous) intensity exercise in the near future?	YES ☐ NO ☐

2021 PAR-Q+

6.	**Do you have any Mental Health Problems or Learning Difficulties?** This includes Alzheimer's, Dementia, Depression, Anxiety Disorder, Eating Disorder, Psychotic Disorder, Intellectual Disability, Down Syndrome If the above condition(s) is/are present, answer questions 6a-6b — If **NO** ☐ go to question 7	
6a.	Do you have difficulty controlling your condition with medications or other physician-prescribed therapies? (Answer **NO** if you are not currently taking medications or other treatments)	**YES** ☐ **NO** ☐
6b.	Do you have Down Syndrome **AND** back problems affecting nerves or muscles?	**YES** ☐ **NO** ☐
7.	**Do you have a Respiratory Disease?** *This includes Chronic Obstructive Pulmonary Disease, Asthma, Pulmonary High Blood Pressure* If the above condition(s) is/are present, answer questions 7a-7d — If **NO** ☐ go to question 8	
7a.	Do you have difficulty controlling your condition with medications or other physician-prescribed therapies? (Answer **NO** if you are not currently taking medications or other treatments)	**YES** ☐ **NO** ☐
7b.	Has your doctor ever said your blood oxygen level is low at rest or during exercise and/or that you require supplemental oxygen therapy?	**YES** ☐ **NO** ☐
7c.	If asthmatic, do you currently have symptoms of chest tightness, wheezing, laboured breathing, consistent cough (more than 2 days/week), or have you used your rescue medication more than twice in the last week?	**YES** ☐ **NO** ☐
7d.	Has your doctor ever said you have high blood pressure in the blood vessels of your lungs?	**YES** ☐ **NO** ☐
8.	**Do you have a Spinal Cord Injury?** *This includes Tetraplegia and Paraplegia* If the above condition(s) is/are present, answer questions 8a-8c — If **NO** ☐ go to question 9	
8a.	Do you have difficulty controlling your condition with medications or other physician-prescribed therapies? (Answer **NO** if you are not currently taking medications or other treatments)	**YES** ☐ **NO** ☐
8b.	Do you commonly exhibit low resting blood pressure significant enough to cause dizziness, light-headedness, and/or fainting?	**YES** ☐ **NO** ☐
8c.	Has your physician indicated that you exhibit sudden bouts of high blood pressure (known as Autonomic Dysreflexia)?	**YES** ☐ **NO** ☐
9.	**Have you had a Stroke?** *This includes Transient Ischemic Attack (TIA) or Cerebrovascular Event* If the above condition(s) is/are present, answer questions 9a-9c — If **NO** ☐ go to question 10	
9a.	Do you have difficulty controlling your condition with medications or other physician-prescribed therapies? (Answer **NO** if you are not currently taking medications or other treatments)	**YES** ☐ **NO** ☐
9b.	Do you have any impairment in walking or mobility?	**YES** ☐ **NO** ☐
9c.	Have you experienced a stroke or impairment in nerves or muscles in the past 6 months?	**YES** ☐ **NO** ☐
10.	**Do you have any other medical condition not listed above or do you have two or more medical conditions?** If you have other medical conditions, answer questions 10a-10c — If **NO** ☐ read the Page 4 recommendations	
10a.	Have you experienced a blackout, fainted, or lost consciousness as a result of a head injury within the last 12 months **OR** have you had a diagnosed concussion within the last 12 months?	**YES** ☐ **NO** ☐
10b.	Do you have a medical condition that is not listed (such as epilepsy, neurological conditions, kidney problems)?	**YES** ☐ **NO** ☐
10c.	Do you currently live with two or more medical conditions?	**YES** ☐ **NO** ☐

PLEASE LIST YOUR MEDICAL CONDITION(S) AND ANY RELATED MEDICATIONS HERE: ______________________

GO to Page 4 for recommendations about your current medical condition(s) and sign the PARTICIPANT DECLARATION.

2021 PAR-Q+

If you answered NO to all of the FOLLOW-UP questions (pgs. 2-3) about your medical condition, you are ready to become more physically active - sign the PARTICIPANT DECLARATION below:

- It is advised that you consult a qualified exercise professional to help you develop a safe and effective physical activity plan to meet your health needs.
- You are encouraged to start slowly and build up gradually - 20 to 60 minutes of low to moderate intensity exercise, 3-5 days per week including aerobic and muscle strengthening exercises.
- As you progress, you should aim to accumulate 150 minutes or more of moderate intensity physical activity per week.
- If you are over the age of 45 yr and **NOT** accustomed to regular vigorous to maximal effort exercise, consult a qualified exercise professional before engaging in this intensity of exercise.

If you answered **YES** to **one or more of the follow-up questions** about your medical condition:
You should seek further information before becoming more physically active or engaging in a fitness appraisal. You should complete the specially designed online screening and exercise recommendations program - the **ePARmed-X+ at www.eparmedx.com** and/or visit a qualified exercise professional to work through the ePARmed-X+ and for further information.

Delay becoming more active if:

- You have a temporary illness such as a cold or fever; it is best to wait until you feel better.
- You are pregnant - talk to your health care practitioner, your physician, a qualified exercise professional, and/or complete the ePARmed-X+ **at www.eparmedx.com** before becoming more physically active.
- Your health changes - talk to your doctor or qualified exercise professional before continuing with any physical activity program.

- You are encouraged to photocopy the PAR-Q+. You must use the entire questionnaire and NO changes are permitted.
- The authors, the PAR-Q+ Collaboration, partner organizations, and their agents assume no liability for persons who undertake physical activity and/or make use of the PAR-Q+ or ePARmed-X+. If in doubt after completing the questionnaire, consult your doctor prior to physical activity.

PARTICIPANT DECLARATION

- All persons who have completed the PAR-Q+ please read and sign the declaration below.
- If you are less than the legal age required for consent or require the assent of a care provider, your parent, guardian or care provider must also sign this form.

I, the undersigned, have read, understood to my full satisfaction and completed this questionnaire. I acknowledge that this physical activity clearance is valid for a maximum of 12 months from the date it is completed and becomes invalid if my condition changes. I also acknowledge that the community/fitness center may retain a copy of this form for records. In these instances, it will maintain the confidentiality of the same, complying with applicable law.

NAME ______________________________ DATE ______________________________

SIGNATURE ______________________________ WITNESS ______________________________

SIGNATURE OF PARENT/GUARDIAN/CARE PROVIDER ______________________________

For more information, please contact
www.eparmedx.com
Email: eparmedx@gmail.com

Citation for PAR-Q+
Warburton DER, Jamnik VK, Bredin SSD, and Gledhill N on behalf of the PAR-Q+ Collaboration. The Physical Activity Readiness Questionnaire for Everyone (PAR-Q+) and Electronic Physical Activity Readiness Medical Examination (ePARmed-X+). Health & Fitness Journal of Canada 4(2):3-23, 2011.

The PAR-Q+ was created using the evidence-based AGREE process (1) by the PAR-Q+ Collaboration chaired by Dr. Darren E. R. Warburton with Dr. Norman Gledhill, Dr. Veronica Jamnik, and Dr. Donald C. McKenzie (2). Production of this document has been made possible through financial contributions from the Public Health Agency of Canada and the BC Ministry of Health Services. The views expressed herein do not necessarily represent the views of the Public Health Agency of Canada or the BC Ministry of Health Services.

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01-11-2020

Appendix III: Physical Activity and Sedentary Behaviour Questionnaire (PASB-Q)

CSEP | SCPE
THE GOLD STANDARD IN EXERCISE SCIENCE AND PERSONAL TRAINING

CSEP-PATH: PHYSICAL ACTIVITY AND SEDENTARY BEHAVIOUR QUESTIONNAIRE (PASB-Q)
ADULT (18 AND OVER)

Please answer the following questions based on what you do in a typical week. To increase accuracy, you may wish to log your physical activity and sedentary behavior for one week prior to answering the questions.

Aerobic Physical Activity

1. Frequency: In a typical week, how many days do you do moderate-intensity (like brisk walking) to vigorous-intensity (like running) aerobic physical activity ?

____ days/week

2. Time or Duration: On average for days that you do at least moderate-intensity aerobic physical activity (as specified above), how many minutes do you do?

____ minutes/day

Total: Multiply your average number of days per week by the average number of minutes per day.

____ minutes/week

Muscle Strengthening Physical Activity

3. In a typical week, how many times do you do muscle strengthening activities (such as resistance training or very heavy gardening)?

____ times/week

Perceived Aerobic Fitness

4. In general, would you say that your aerobic fitness (ability to walk/run distances) is:

____ Excellent ____ Very Good ____ Good ____ Fair ____ Poor

Sedentary Behaviour

5. On a typical day, how many hours do you spend in continuous sitting: at work, in meetings, volunteer commitments and commuting (i.e., by motorized transport)?

☐ None	☐ < 1 hour	☐ 1 to < 2	☐ 2 to < 3
☐ 3 to < 4	☐ 4 to < 5	☐ 5 to < 6	☐ > 6

6. On a typical day, how many hours do you watch television, use a computer, read, and spend sitting quietly during your leisure time?

☐ None	☐ < 1 hour	☐ 1 to < 2	☐ 2 to < 3
☐ 3 to < 4	☐ 4 to < 5	☐ 5 to < 6	☐ > 6

Total Sedentary Behaviour (add responses to questions 5 and 6)____ hours/day

7. When sitting for prolonged periods (one hour or more), at what interval would you typically take a break to stand and move around for two minutes?

☐ < 10 minutes

☐ 10 to < 20 minutes

☐ 20 to < 30 minutes

☐ 30 to < 45 minutes

☐ 45 to < 1 hour

☐ 1 to < 1.5 hours

☐ 1.5 to < 2 hours

☐ > 2 hours

Appendix IV: 12-Month Fall History and Falls Efficacy Scale-International (FES-I)

In the <u>past year</u>, have you had any fall including a slip or trip in which you lost your balance and landed on the floor or ground or lower level?

Yes [ ] No [ ]

If you have had <u>no falls</u> please skip straight to section 4, otherwise please continue.

If yes, how many times has this happened?

Once [ ] Twice [ ] Three or more [ ]

1. Where have you fallen?

<u>Inside:</u>

On the one ground level	Yes [ ]	No [ ]
Getting out of bed	Yes [ ]	No [ ]
Getting out of a chair	Yes [ ]	No [ ]
Using the shower/bath	Yes [ ]	No [ ]
Using the toilet	Yes [ ]	No [ ]
Walking up or down stairs	Yes [ ]	No [ ]

<u>Home entrances or in the garden:</u>

Walking up or down a step/stairs	Yes [ ]	No [ ]
On the one ground level (e.g. pathway)	Yes [ ]	No [ ]
In the garden	Yes [ ]	No [ ]

<u>Away from home:</u>

On a footpath	Yes [ ]	No [ ]
On a kerb/gutter	Yes [ ]	No [ ]
In a public building	Yes [ ]	No [ ]
Getting out of a vehicle	Yes [ ]	No [ ]
In another person's home	Yes [ ]	No [ ]

Other (please describe):

2. How did you fall?

I tripped	[ ]
I slipped	[ ]
I lost my balance	[ ]
My legs gave way	[ ]
I felt faint	[ ]
I felt giddy/dizzy	[ ]
I am not sure	[ ]

3. As a result of this fall or falls did you suffer any injuries?

Yes [ ] No [ ]

If yes, what type of injuries did you suffer?

Bruises	[ ]
Cuts/grazes	[ ]
Broken wrist	[ ]
Broken hip	[ ]
Broken ribs	[ ]
Back pain	[ ]

Other (please specify):

4. Now we would like to ask some questions about how concerned you are about the possibility of falling. Please reply thinking about how you usually do the activity. If you currently don't do the activity (e.g. if someone does your shopping for you), please answer to show whether you think you would be concerned about falling IF you did the activity. For each of the following activities, please tick the box which is closest to your own opinion to show how concerned you are that you might fall if you did this activity.

		Not at all concerned 1	*Somewhat concerned 2*	*Fairly concerned 3*	*Very concerned 4*
1	Cleaning the house (e.g. sweep, vacuum or dust)	*1 [ ]*	*2 [ ]*	*3 [ ]*	*4 [ ]*
2	Getting dressed or undressed	*1 [ ]*	*2 [ ]*	*3 [ ]*	*4 [ ]*
3	Preparing simple meals	*1 [ ]*	*2 [ ]*	*3 [ ]*	*4 [ ]*
4	Taking a bath or shower	*1 [ ]*	*2 [ ]*	*3 [ ]*	*4 [ ]*
5	Going to the shop	*1 [ ]*	*2 [ ]*	*3 [ ]*	*4 [ ]*
6	Getting in or out of a chair	*1 [ ]*	*2 [ ]*	*3 [ ]*	*4 [ ]*
7	Going up or down stairs	*1 [ ]*	*2 [ ]*	*3 [ ]*	*4 [ ]*
8	Walking around in the neighbourhood	*1 [ ]*	*2 [ ]*	*3 [ ]*	*4 [ ]*
9	Reaching for something above your head or on the ground	*1 [ ]*	*2 [ ]*	*3 [ ]*	*4 [ ]*
10	Going to answer the telephone before it stops ringing	*1 [ ]*	*2 [ ]*	*3 [ ]*	*4 [ ]*
11	Walking on a slippery surface (e.g. wet or icy)	*1 [ ]*	*2 [ ]*	*3 [ ]*	*4 [ ]*
12	Visiting a friend or relative	*1 [ ]*	*2 [ ]*	*3 [ ]*	*4 [ ]*
13	Walking in a place with crowds	*1 [ ]*	*2 [ ]*	*3 [ ]*	*4 [ ]*
14	Walking on an uneven surface (e.g. rocky ground, poorly maintained pavement)	*1 [ ]*	*2 [ ]*	*3 [ ]*	*4 [ ]*
15	Walking up or down a slope	*1 [ ]*	*2 [ ]*	*3 [ ]*	*4 [ ]*
16	Going out to a social event (e.g. religious service, family gathering or club meeting)	*1 [ ]*	*2 [ ]*	*3 [ ]*	*4 [ ]*