The Age Related Differences in Individual Joint Contributions of the Ankle, Knee and Hip in Forward Compensatory Stepping

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Abstract

Problem: Falls in older adults are prevalent in Canada which cause physical, psychological, emotional and financial stressors. With an increasingly older adult population, fall prevention is key in solving this problem. Learning how to prevent falls starts by understanding the difference(s) between older adults and younger adults in balance recovery mechanisms.

Purpose: The purpose of this study is to investigate whether there were any age-related differences in stepping-lib joint moments and powers during the execution of a single-step compensatory balance response.

Methods: A lean and release protocol was utilized to apply a balance disturbance to the participants. Participants were tasked to recover from the perturbation and restore stability using a single step with their preferred leg. The peak moment magnitude, peak moment power and timings of these peaks were recorded at the ankle, knee and hip in both the sagittal and frontal planes. Independent samples t-tests were used to compare joint-specific variables between older adults and younger adults to understand if age-related differences in stability may arise from challenges in moment generation or power absorption at specific joints.

Results: Our results showed age-related differences in peak ankle moments in the sagittal plane as well as peak ankle power magnitudes in both the sagittal and frontal plane. In addition to these findings, our results also found key age-related differences in the timing of peak moments in the sagittal plane with both the hip and the knee joints as well as differences in the timing of peak power values in both the sagittal plane and frontal plane at the ankle, knee and hip joints.

Conclusion: Age-related differences in the execution of the single-step compensatory response were found between older adults and younger adults. Our joint-specific findings can be used to further our understanding of balance control strategies in older adults but it also highlights the importance of timings in peak moment and power at the ankle, knee and hip which can be used to improve fall prevention and fall rehabilitation programs in older adults.

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INTRODUCTION

Canadians are very blessed to live in one of many countries that has publicly funded healthcare. This privilege can, however, put a large financial burden on our economy. In 2004, Canada spent an estimated $19.8 billion due to unintentional injuries and this has been the leading cause of hospitalization in children, young adults and older adults. Not only that, this is also the major cause of disability and death in Canada (http://parachutecanada.org). Among the top causes of injuries are falls, which continue to be the leading cause of injury-related hospitalizations among seniors in Canada. Out of the total $19.8 billion spent on unintentional injuries, $2 billion was spent as direct costs associated with falls among older adults (Public Health Agency of Canada, 2014).

Falls are prevalent among older adults; the Public Health Agency of Canada (2014) has estimated that between 20%-30% of older adults fall each year. In addition, Statistics Canada (2011) has estimated that there are five million Canadians who are over the age of 65, which constitutes 15% of the total population. As the Baby Boomer Generation continue to age, the number of older adults is expected to increase. It is expected that in the next 25 years, the number of older adults will double, which means that there will be an estimated 10 million adults, over the age of 65 years, by the year 2036 (Public Health Agency of Canada, 2014). Due to this shift towards an older demographic, there is also an expectation for the absolute number of falls to increase if action is not taken to reduce falls. With the number of deaths due to falls increasing every year (Statistics Canada, 2012), it is imperative to handle the situation urgently to find ways to reduce the number of falls especially in the older population. Since these are injury-related incidences, there are definite ways to prevent and reduce such harm. The first step to solving this problem is to have an in-depth understanding of falls in terms of how and why they occur.

Falls are often defined as “a sudden and unintentional change in position resulting in an individual landing at a lower level such as on an object, the floor, or the ground, with or without [sustaining an] injury” (Tinetti et al., 1997). These physical injuries include a wide variety ranging from simple abrasions to fractures or sometimes, even death. In addition to physical consequences, falls can also have negative mental health consequences such as fear of falling, loss of autonomy, isolation, confusion, immobilization and depression (Public Health Agency of Canada, 2014). Out of all these injuries, the main injury that is directly associated with falls are hip/pelvic fractures. Ninety-five percent of hip fractures are due to falls and this leads to death in 20% of all hip fracture cases (Ioannidis et al., 2009, Jiang et al., 2005, and Wolinksky et al., 2009).

Falls are initiated when an individual loses their balance. Balance can be classified as either static or dynamic. Static balance is the ability to maintain an upright posture in a fixed position while dynamic balance is the ability to maintain an upright posture while the body is in motion. Static balance is maintained when the centre of mass (CoM) is found within the body’s base of support (BoS) (O’Sullivan & Portnry, 2014). When the body’s CoM is not within the body’s BoS due to an internal or external perturbation, it results in a loss of balance. Fortunately, our bodies have a range of innate strategies to restore our balance following a perturbation to prevent falls from occurring such as the hip strategy, the ankle strategy (Horak & Nasher, 1986), and the compensatory step (McIlroy & Maki, 1993). Over the years, the compensatory step has been studied yet, despite similarities between older and younger adults in the initial temporospatial aspects of the response, older adults have been found to exhibit continued or additional instability after the initial response has been executed. As such, new work is beginning to examine the restabilisation phase, following foot contact in compensatory stepping, as a vital phase for balance recovery and fall prevention (Singer et al., 2014, 2016, King et al. 2011, Weerdesteyn et al., 2011).

Compensatory stepping is a mechanism used by our body as a means to increase the BoS and the external moment of force about the CoM to achieve a greater likelihood of restabilising the body from a given perturbation. This involves a step(s) in the direction of the body's CoM velocity as an attempt to regain static balance by increasing the BoS to accommodate for the body's change in CoM (Singer, Prentice & McIlroy, 2016). Due to this step(s), this mechanism is considered a change-in-support strategy rather than a fixed-support strategy such as the ankle and hip strategies (Maki & McIlroy, 1997). The restabilisation phase of compensatory stepping begins immediately after foot contact until a successful balance recovery is achieved. It is in this phase where the individual has the ability to control the body's centre of pressure (CoP) as well as the magnitude and the line of action of the ground reaction forces (GRF), which are important for restabilisation as these components influence whole-body linear and angular accelerations (Singer et al., 2014). Age-related differences have been identified during the restabilisation phase of the compensatory stepping response (Rogers et al., 2001, McIlroy & Maki, 1996) and might be associated with the increased prevalence of instability and falls in older adults in comparison to younger adults.

In previous studies, some significant differences have already been identified in the execution of the initial phases of the compensatory step between younger adults (YA) and older adults (OA). For example, OA displayed an increased peak in mediolateral GRFs, from beneath the stepping limb, compared to the YA (King et al., 2011). Differences were also found between OA and YA in terms of muscle activation timing and maximum response execution speed during swing phase of the compensatory step despite not having any differences in response onset following the initial perturbation (Thelen et al., 1997,2000). In addition, a study by Oberg et al. (1993) revealed that OA, when recovering balance by compensatory stepping, have a reduction in not only stepping speed but also in step length. Oberg et al. (1993) also found that YA produced larger ankle plantarflexor and hip extensor moments than OA at the initiation phase of compensatory stepping which resulted in shorter contact times of the foot to the ground. It is due to this that YA are associated to being able to recover balance from a larger perturbation than OA.

Studies examining the restabilisation phase (following step contact) during submaximal perturbations, which may be experienced during activities of daily living, have focused on understanding the control of applied forces by the lower limbs, which directly dictate CoM kinematics. Previous work has examined the eccentricity of the net ground reaction force (GRF_{net}) relative to the whole-body CoM, which regulates whole-body linear and angular momentum. The eccentricity of the GRF_{net} was quantified as the angle of divergence, which is the deviation angle measured from the line between the body's CoM and the CoP_{net}—a zero-angle denotes a centric force while a non-zero angle is considered to be an eccentric force. Two peak GRF_{net} eccentricities (termed P1 and P2) emerged in both sagittal and frontal planes and have each been linked with anteroposterior and mediolateral stability control, respectively. The P1 eccentricity emerges at approximately 50ms following step contact and is thought to be mediated proactively by preparatory muscle activation and limb stiffness on foot contact—the magnitude of the sagittal plane P1 is associated with A/P stability. The P2 eccentricity emerges at approximately 200ms following step-contact and is thought to be modulated reactively in response to instability that persists or arises during the restabilisation phase—the timing of P2 in the mediolateral direction has been associated with M/L stability (Singer et al., 2014, 2016, 2019). Although we now have understanding of how the whole-body kinetic restabilisation response (i.e. characteristics of the GRF_{net}) differs between older and younger adults, we still lack specific joint-level information that may be used as foci for balance interventions. With better information about the role of individual lower limb joints in regulating force output during restabilisation from an external balance perturbation, it may be possible to develop more targeted balance exercises that address an individual's specific balance control challenges. The current study seeks to further understand the mechanisms involved in stability during the restabilisation phase of compensatory stepping by investigating age-related differences in ankle, knee and hip moments of force, intersegmental powers and timings of moments and powers. With such understanding, we may be able to further the development of exercise-based balance intervention programs by tailoring these interventions to the specific control challenges identified.

LITERATURE REVIEW

Falls in Older Adults

Falls are the leading cause of injury-related hospitalizations among older adults in Canada (Canadian Institute for Health Information, 2010). According to Statistics Canada (2010), 20% of older adults living in the community reported a fall. Among these, there was a higher prevalence of individuals over the age of 80 years old. In 2009 and 2010, the highest prevalence of older adults falling was between ages 85-89 years old (Statistics Canada, 2012). Even though a lot of falls occur in the winter where snow and ice are present on the ground, the largest percentage of falls happen inside one's own residence, which comprises 50% of all fall-related hospitalizations, and in residential institutions, which comprises 17% of all fall-related hospitalizations (Public Health Agency of Canada, 2014).

Falls can result in a wide variety of injuries throughout the body. According to Statistics Canada (2012), the greatest number of injuries sustained as a result from falls (in older adults over the age of 65) were shoulder or upper arm injuries which constitute 17% of all injuries followed by knee or lower leg injuries at 15%. Ten percent of injuries occurred at the foot or ankle and 8% occurred at the lower back or spine. The remaining injuries were: 8% at the wrist, 7% at the hip, 7% head injuries (including facial bones), 7% at the chest, 4% at the hand, 4% at the elbow or lower arm, 4% reported injuries on multiple body parts, 3% at the abdomen or pelvis, 2% at the thigh, 3% at the neck or upper back or upper spine and the remaining 1% were injuries to the eyes. Statistics Canada (2012) also outlined the types of fall-related injuries sustained in older adults over the age of 65. The largest number of injuries sustained as a result from falls were fractures which constitute 35% of all types of injuries followed by sprains or strains at 30%. The remaining types or injuries were: scrapes, bruises or blisters at 19%, cuts, punctures or bites at 6%, dislocations at 3%, concussions or other brain injuries at 2% and “other” for the remaining 5%. Despite the wide variety of injuries that can be sustained from a fall, hip fractures were directly associated with falls. 95% of all hip fractures were due to falls; death occurred in 20% of all hip fracture patients (Ioannidis et al., 2009, Jiang et al., 2005, and Wolinksky et al., 2009). Due to the large social and economic impact of fall-related fractures, better understanding of the mechanisms related to instability may be helpful to improve components of fall-prevention programs.

Balance Control Strategies

Falls can be devastating due to the wide range of injuries that can be sustained. Fortunately, there exist multiple balance mechanisms/strategies to attempt to restore balance after a balance perturbation. These strategies are composed of centrally organized muscle activation patterns that respond to initial conditions, perturbation characteristics, learning and intention (Horak & Macpherson, 1996, Macpherson, 1991, Lee, 1984). These discovered strategies are “not hardwired like reflexes but [instead, can] be gradually learned with experience in new environmental contexts” (Horak & Nashner, 1986). In addition, Horak et al. (1997) also found that these learned strategies are capable of becoming more efficient with practice. According to a number of studies, the chosen muscle synergies used by our bodies to respond to an external perturbation depend on a few factors such as initial body position (Macpherson et al., 1989, Horak & Moore, 1993), initial support conditions (Nashner & Cordo, 1981, Schieppati & Nardone, 1991), and the location and characteristics of the sensory stimuli that triggered the response (Diener et al., 1988). Any impairments to detecting these factors will affect the execution of the strategy in recovering balance from a perturbation.

Among such postural response strategies are the ankle strategy and the hip strategy which are used to maintain equilibrium following a disturbance of balance of varying magnitudes in the anterior or posterior position (Nashner & McCollum, 1985, Horak & Nashner, 1986, Horak et al., 1990). These strategies are considered to be fixed-support strategies which means that there is no change in the body's BoS and are named after the joint at which the most movement occurs in the body, which also corresponds to the manner of how the nervous system restores balance through the generation of joint moments of force (Rietdyk et al., 1999). The ankle strategy uses a distal-to-proximal muscle activation pattern to move the body's CoM with torques primarily at the ankle and the knee joints while the hip strategy uses early proximal hip and trunk muscle activation to generate hip torque in addition to the ankle and knee torque present (Horak et al., 1997). Despite the distinct features between the ankle strategy and the hip strategy, early studies have shown that these two strategies are on the same continuum of locomotor strategies (Nashner, 1977) and can be combined to provide a mixed strategy response to achieve equilibrium after a perturbation (Horak & Nashner, 1986).

According to the inverted pendulum model proposed by Winter et al. (1998), the centre of pressure (CoP) is the variable used to control the body's CoM. The ankle strategy and the hip strategy both influence a body's balance through the control of the body's CoP. In the ankle strategy, Horak & Nashner (1986) observed that there is a torsional resistance at the ankles that coincide with the acceleration of the body's CoM back toward the position of equilibrium. Similarly, in the hip strategy, a large active rotation occurs at the hip joint to generate shear forces at the feet to decelerate the CoM.

Aside from the fixed-support strategies, there are also balance control mechanisms called change-in-support strategies, which from their name, involve a change in the BoS of an individual to maintain balance. Unlike the fixed-support strategies where movements can be combined to elicit a mixed strategy, change-in-support strategies are independent mechanisms and are usually the primary mechanisms used compared to the fixed-support strategies when there are no surface or instructional constraints (e.g. specifically given instructions not to take a step to recover balance) (Horak et al., 1997). Change-in-support strategies are crucial in balance recovery as they are the only reactions that can successfully maintain balance after a large perturbation. However, this does not mean that change-in-support strategies are only last resort mechanisms. According to a few studies (Maki & Whitelaw, 1993, Maki et al., 1993, McIlroy & Maki, 1993, McIlroy & Maki, 1995), compensatory limb movements are common reactions to external postural perturbations even when the magnitude of the disturbances are small and stability could have been maintained without moving the arms or legs. In addition, a video surveillance study (Holliday et al., 1990) also showed that compensatory limb movements are common reactions to balance disturbances in daily life with compensatory stepping performed in 32%-45% of falls or near-falls and arm movements performed in 65%-72% of all incidences.

Change-in-support strategies restore balance, following a perturbation, by increasing the size of the BoS in order to accommodate the CoM displacement without the loss of stability. These strategies also increase the moment arm between the point of force application of either the foot (in compensatory stepping) or hand (in compensatory grasping) and the CoM. As a result, it greatly amplifies the stabilizing moments which act to decelerate the CoM. In addition, grasping reactions can also decelerate the CoM by serving as an anchor for the body to a supportive structure. However, despite the huge benefits that change-in-support mechanisms offer, inaccurate or inappropriately timed or scaled limb movements may add destabilizing forces and moments which can further lead to falls (Maki & McIlroy, 1997). Due to this, it is imperative to understand the mechanisms behind these strategies in order to execute these skills properly and reap their benefits.

Among these change-in-support strategies are compensatory grasping and compensatory stepping. According to Maki & McIlroy (1997), compensatory grasping is a central nervous system (CNS) driven ability that “rapidly and accurately controls the trajectory of the hand to a fixed [stable] target” to maintain balance during a perturbation. Not only are these movements used to restore balance, the arm movements involved also serve as a protective role to shield the head from possible impact. Although not a lot of studies have been performed focusing on compensatory grasping, a study by McIlroy & Maki (1995) observed a unique feature of compensatory grasping in which the arm responses used occurred without any prior stretch nor loading of the arm muscles. Similarly, a study by Maki & McIlroy (1997) also observed that both the arm and the shoulder muscles only became active in maintaining balance after the perturbation has already occurred. They also observed a high frequency (85% of all trials in their study) of compensatory grasping responses occurring even when the participants were not given specific instructions. As a result, Maki & McIlroy (1997) conclude that compensatory grasping is not simply just a startle response to balance perturbation but a highly sophisticated ability controlled by the CNS which involves rapid initiation and amplitude scaling of hand trajectories to attempt to restore balance. The neural pathways used to execute compensatory grasping have been found to be controlled by transcortical or subcortical pathways, which are similar pathways used in the early stages of fixed-support postural control mechanisms.

Compensatory Stepping

The overall focus of our study involves the lower limb change-in-support strategy called the compensatory stepping response. This response involves a step to recover balance from a perturbation by changing the BoS to accommodate for the change in CoM dynamics. Although there are numerous studies conducted focusing on the compensatory step, there is very little known about how this response is controlled by the CNS. Even with our limited understanding, sensory feedback is known to play an important role in compensatory stepping. A study by Perry et al. (2000) involving the plantar pressure feedback from the soles of our feet indicates that the loss in plantar sensation increases the incidence of multi-step responses to recover balance from a perturbation. This multi-stepping response often reflects the impaired ability of the individual to control the lateral displacement of the CoM during compensatory stepping (Maki & McIlroy, 1997) and is an indicator of lateral instability. In addition, another study (McIlroy et al., unpublished research) involving the full loss of sensation of the sole of the foot due to anesthesia found that the participants were unable to maintain balance during a single-leg stance using the affected foot. With these findings, afferent information is found to be vital in controlling the swing limb unloading, foot lifting, foot contact and weight transfer. However, apart from the sensory information obtained from the feet, interactions between the vestibular and visual systems have also been found to play an important part in initiating the step response (Hoshiyama et al., 1993). A concrete example of the integration between the visual and vestibular systems is when a person moves forward (e.g. walking), the apparent movement of the entire visual field (optic flow) moves in the opposite direction to the displacement of the body. As a result of these visual cues, even when there is no displacement of the body, but there is still movement of the visual field, this still causes a sensation that the body is being displaced. Research by Allum et al. (1976) have discovered that visual information such as visual flow interacts with the information from the vestibular and proprioceptive systems in a reflex-like reaction. Due to the short latency times of these interactions, it is important to also acknowledge these mechanisms involved in studying the compensatory step.

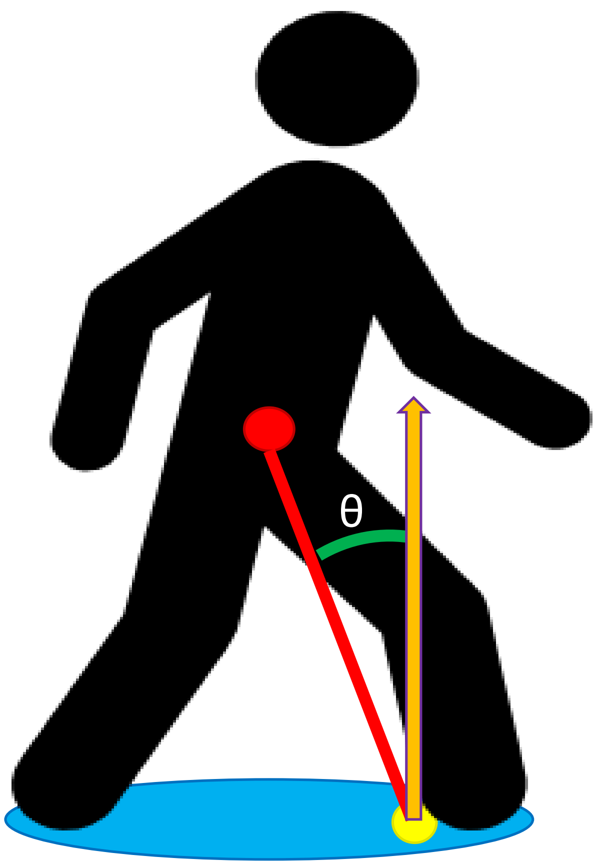
The compensatory stepping response is not to be confused with non-compensatory or volitional stepping as seen in gait initiation because both mechanisms show fundamental differences when compared to each other. One of these key differences is the speed of response. Previous studies (McIlroy & Maki, 1993, Burleigh et al., 1994) have compared the speed of response between volitional and compensatory stepping using different protocols and the findings are the same—compensatory stepping response time is faster than volitional stepping. The rapid response can also be witnessed even when the participants are instructed not to step, however, a few participants were able to delay the response. This delay was found to be more common in forward compensatory stepping versus backward compensatory stepping probably due to the ability to create stabilizing torques by bringing the CoP forward toward the toes through the utilization of fixed-support strategies. Even when the response is delayed though, Maki et al. (1996) observed that the speed at which the swing leg is unloaded is still extremely fast during compensatory stepping in comparison to volitional stepping.

Another fundamental difference between compensatory stepping and volitional stepping is the absence of anticipatory postural adjustments (APA) in compensatory stepping reactions (Maki & McIlroy, 1997, Rogers, 1996, McIlroy & Maki, 1993, 1995, 1999), which allows for a faster swing limb loading time. These APAs have both anterior/posterior (A/P) direction and mediolateral (M/L) components. A/P APAs involve the shifting of body weight to the swing leg first then back to the stance leg prior to the lifting of the swing leg. Meanwhile, in M/L APAs the CoM is moved toward the stance limb to promote stability by reducing the tendency of the CoM to fall towards the unsupported side during subsequent single support in swing phase (Breniere et al., 1987, Jian et al., 1994). This shifting of body weight, and displacement of the CoM toward the stance limb, results in improved mediolateral stability in an upright posture during leg flexion (Rogers, 1922, Rogers & Pai, 1990, Do et al., 1991), leg abduction (Mouchnino et al., 1992) and gait initiation (Mann et al., 1979, Breniere et al., 1987, Nissan & Whittle, 1990, Brunt et al., 1991, Jian et al., 1994). The absence of APAs in compensatory stepping suggest that this could be due to the absence of preplanning.

Studies (McIlroy & Maki, 1993, 1996, Burleigh et al., 1994, Burleigh & Horak, 1996) have shown that M/L APAs occur more consistently when participants are given prior instructions to step in response to the balance perturbation. In addition, studies (McIlroy & Maki, 1993, 1995, 1996, Rogers, 1996) have also found that APAs are absent during early trials when the perturbation is unfamiliar and when participants were not given prior instructions to the step. Although APAs help in maintaining balance, they also come with a cost that could jeopardize balance—that cost is a delay in response. With a delayed response after the perturbation, the participant's CoM spends a greater period of time falling laterally, which exposes the participant to an unstable position for a longer period of time. According to McIlroy & Maki (1995), there is about a 100 millisecond delay in the lifting and placement of the swing foot in compensatory stepping if an APA is included. Liu et al. (2003) also noticed that the amplitude of the APA also has an effect on the latency of foot lifting. Larger amplitude APAs were associated with longer latency of foot lifting where latency of foot lifting is defined as the time period from the onset of the perturbation to the time when the swing foot lifts off the ground.

The biomechanical features of compensatory stepping focus on the kinematic and kinetic measures that occur throughout the entire process. When a large perturbation is experienced by the body, the body's CoM is displaced. The compensatory stepping mechanism can restore the body's dynamic balance by moving the foot to increase the body's BoS to accommodate for the CoM displacement. The net ground reaction force (GRF_{net}) that is generated at foot contact is used to create a stabilizing moment to counteract the destabilizing moments due to the perturbation as well as the transition from a single leg stance to a double-leg stance. In studies by Singer et al. (2014, 2016, 2019), a measure called the angle of divergence was introduced to simplify the relationship between the GRF and the position of the body's CoM. The angle of divergence is the angle formed between the body's CoM, the foot CoP and the orientation of the GRF. A positive angle of divergence, where the GRF_{net} is anterior or lateral to the CoM, represents a stabilizing moment meanwhile a negative angle of divergence, where the GRF_{net} is posterior or medial to the CoM, represents a destabilizing moment (Figure 1).

Figure 1: Angle of Divergence.



Legend:

Θ = Angle of divergence

Note: If the orientation of GRF_{net} is anterior or lateral to the COM, it creates a positive value while a negative value is created when the GRF_{net} is posterior or medial to the CoM.

Red line = CoM-CoP inclination angle

Orange line = Orientation of GRF_{net}

Yellow circle = Foot CoP

There have been previous studies (King et al., 2011; Kurz et al., 2013; Singer et al., 2014, 2016; Weerdesteyn et al., 2011) that highlight the importance of the restabilisation phase—which starts at the moment of foot contact up until balance has been fully restored. It is in this phase where the body is given the opportunity to change body kinematics by controlling the CoP, together with the magnitude and orientation of the GRFs, to restore dynamic stability. Within this phase, two distinct time points have been identified in which the angle of divergence peaks in such a way to enable whole-body restabilisation. Due to their major contribution to the recovery of balance, these two peaks represent the net stabilizing effect of the GRFs in the entire restabilisation phase. The first peak, P1, occurs in a window between the point of foot-contact and within 100ms after foot-contact. The onset of this peak is considered to be very fast and due to this, it is not considered to be modulated reactively. On the other hand, the second peak, P2, which occurs later, is found to occur fairly consistently at 200ms following foot contact during both compensatory stepping (Singer et al., 2014, 2016, 2019) and steady-state gait (Rawal & Singer, 2021). Due to the latency at which P2 occurs, coupled with similarity to lower limb force output during platform perturbations (Horak et al., 1997), it is considered to be reactively modulated in response to instability following foot contact. Due to the potential for P2 to be governed by reactive control during restabilisation, its characteristics may be key to optimizing the body's compensatory stepping response (Singer et al., 2016). The regulation of the orientation of the GRF plays an important role in whole-body angular momentum control and, consequently, as a determinant of dynamic stability as seen in gait studies (Herr &

Popovic, 2008; Simoneau & Krebs, 2000; Rawal et al., 2021) and balance recovery after tripping (Pijnapples et al., 2004).

Age-related Differences in Compensatory Stepping

Due to the high incidence of severe injuries to older adults due to falls, much research has been done examining younger adults and older adults to identify similarities and/or differences on the execution of compensatory stepping to recover balance after a perturbation. Over the years, different protocols have been developed to study the age-related differences in compensatory stepping between younger and older adults. A few of these notable studies and their findings will be discussed.

One of the first studies that investigated the age-related differences in compensatory stepping was conducted by McIlroy & Maki (1996). In their experiment, the investigators utilized a surface perturbation protocol using a movable platform. The platform would deliver a perturbation in the A/P direction and the participants would have to recover their balance using the compensatory stepping response. According to McIlroy & Maki (1996), the absolute strength differences found between younger and older adults as an age-related decline was not a major factor in compensatory stepping due to the submaximal effort needed to recover from these perturbations. However, the age-related decline in sensory inputs may cause an increase in conduction time or central processing time of the given response. In this study, older adults were observed to opt into using a multiple stepping strategy when performing the compensatory stepping response meanwhile the younger adults executed the response using only a single step. This multiple stepping strategy utilized by the OA might be a conservative strategy which allows for more opportunities to correct for the instability. Furthermore, the multiple stepping seems to emerge as a result of the events after the initiation of the first step rather than a preplanned strategy, which further supports the idea that the challenges that older adults experience may occur after foot-contact. Consistent with this idea, McIlroy & Maki (1996) observed multiple similarities between the younger and older adults in the execution of the compensatory stepping prior to the restabilisation phase. Among these similarities were: the speed of movement, swing durations, swing velocity, step length, time to loading and unloading of the swing leg, time to foot lift off, and onset of M/L asymmetry. In addition, the presence of an APA was also found to be too small to have an effect on the CoM lateral displacement of the participants. Overall, McIlroy & Maki (1996) concluded that the major issues that older adults face in recovering balance after a perturbation are due to the poor detection of the perturbation as well as a poor response to the perturbation, potentially after foot-contact.

Thelen et al. (1997) also studied the age-related differences in compensatory forward stepping but utilized a lean-release protocol instead of a support-surface perturbation protocol as used by McIlroy & Maki (1996). A lean-release protocol involves suspending participants by a cable that is attached to their back while the participant leans forward (with both feet flat on the ground) to a certain angle based on the desired perturbation. Once in position, the supporting cable is then released and the participant falls and recovers their balance using the compensatory stepping response. The release time of the cable was done at random to minimize any APAs which could influence the execution of the compensatory step. In this study by Thelen et al. (1997), the maximum lean angles at which older adults and younger adults could successfully recover from were compared. It was noted that the younger adults were able to recover their balance from much greater lean angles compared to the older adults. It is also worth noting that older adults, similar to what was observed in the study by McIlroy & Maki (1996), were observed to take multiple steps in recovering balance after a perturbation. Despite these differences that have been identified, both YA and OA exhibited similar levels of motivation in performing the given task and the recorded reaction times were also very similar.

A few years later, a study by Thelen et al. (2000) was released which built on their previous study in 1997 regarding the compensatory stepping mechanism. However, this time around, the focus of their study was to investigate the presence and influence of muscle activation delays and muscle activation timing during the compensatory step. Using the lean-release protocol, the investigators noticed that older adults tend to rely on joint stiffness more than younger adults in recovering balance, based on their increased activity of antagonistic muscles used. Other differences identified within this study include observations made when both younger and older adults were exposed to maximal leans/perturbations. At maximum lean angles, there is a decrease in step length and step velocity noted, as well as an increase in muscular delay found in older adults in comparison to younger adults. Again, despite these differences outlined, numerous similarities were found as well: A.) Step timing and step length between younger and older adults are found to be similar in submaximal perturbations. B.) Both groups also displayed similar preparation phases prior to the execution of the compensatory step. C.) Myoelectric latency measures were similar between older and younger adults. In both studies (Thelen et al., 1997, 2000), the age-related differences in muscle behavior were all found during the stepping movement (prior to and during swing phase) and not due to the slowness in the initiation of the compensatory stepping response.

Rogers et al. (2001) took a different approach and utilized a waist-perturbation protocol to examine the compensatory step. This method involves a waist band that is wrapped around the participant which is connected to a motor-driven machine that applies the desired perturbation to the waist band via a flexible cable. In this study, the authors focused on the position of the CoM throughout the execution of the compensatory step especially in the M/L direction. The measure of M/L stability has been found to be the best predictor of falls (Maki et al., 1994) which is an important factor to consider since falls are a major concern in older adults. In the experiment, older adults were found to have an increased lateral body motion at first step foot contact which was associated with a more lateral foot placement compared to younger adults. This difference in the amount of lateral body motion in older adults versus younger adults was not attributed to any changes/differences in APAs. In addition to these findings, the OA, who had experienced a fall within 12 months prior to the study, exhibited an increase in step duration time compared to the younger adults. According to the authors, this increase in step duration is critical because this means that the body will spend a longer time in single-limb support which allows for a greater period of time where the participants fall laterally at a greater velocity. Altogether, the study by Rogers et al. (2001) highlights the association between instability in older adults and their challenges in controlling lateral body motion during compensatory stepping.

By 2011, many variations of different protocols had been performed with different researchers finding age-related differences (such as multi-stepping response) and age-related similarities (such as step length, swing speed and etc.) in the use of compensatory stepping to recover balance after a perturbation. Later that year, a study by King et al. (2011) was published which closely examined both the kinematic and kinetic measures involved in compensatory stepping. The investigators used a lean-release protocol to conduct their study and found that the kinematic measures such as step width, step clearance and step duration, were similar between younger and older adults when performing compensatory stepping to recover balance after a perturbation. Aside from the kinematic measures, the authors also observed that both younger and older adults follow very similar preparatory phases and utilize very similar APA patterns. Although the kinematic measures were very similar in both age groups, step length was the only measure that showed an age-related difference in this study. The authors believed that this difference might be due to the single step restriction that was imposed in the study. As for the kinetic measures observed, older adults were found to exhibit an increase in kinetic measures of lateral stability. This can be seen in the increase of GRF and ankle forces that older adults utilized to recover balance using a single step. Due to this, it is known that older adults increase strength usage when performing the compensatory step compared to younger adults. In summary, the researchers suggest that “older adults preserve lateral stability through adaptations to lateral kinetic characteristics of stepping” (King et al., 2011). In addition, the researchers also concluded that the age-related differences in the compensatory stepping were not kinematic measures but were kinetic measures of lateral stability. It remains to be known, however, that OA exhibit kinetic modifications in joints proximal to the ankle, which may account for age related differences in lateral stability.

The most recent research in compensatory stepping that was used to provide as an anchor point for the current experiment begins with the study by Singer et al. (2013). This study focused on the restabilisation phase in voluntary movement termination. It served as a foundation that identifies the importance of the restabilisation phase, after foot contact. In this study, the difference between the peak CoM position and the final CoM position (termed CoM incongruity) was used to determine the level of stability—a high CoM incongruity measure suggests that the individual is more unstable, as the CoM is allowed to travel a greater distance laterally or anteriorly, toward the limits of the base of support, before it is brought to rest. In majority of the trials, the participants displayed overshooting of the final CoM position, which suggests that their CoM travels too far laterally or anteriorly before coming to rest within the base of support. As such, it is believed the overshooting that occurs from recovering balance during a perturbation can cause further instability. Overshooting was also purported to arise from challenges in generating an external moment of force about the whole-body CoM, which can be helpful in attempting to regain stability after a perturbation by reversing the CoM angular velocity.

These findings were further supported in the next study by Singer et al. (2014) which focused on M/L kinetic mechanisms involved in restabilisation during voluntary stepping. In addition, the study also did a comparison between younger and older adults to identify if there are any age-related differences in the execution of the task. Age-related differences were identified in this study as older adults displayed an increase in CoM incongruity and an increase in trial-to-trial variability of CoM incongruity compared to the younger adults. With these results, it suggests that older adults may possess difficulties in regulating the magnitude, timing and direction of the GRF to produce a stabilizing moment during the restabilisation phase. Despite these differences, CoM incongruity was also found in younger adults but not to the same magnitude as found in older adults. The study supports the importance of overshooting in stability during movement termination as outlined in the previous study. Despite the age-related differences discovered in this study, similarities were also found in this study between how older and younger adults execute the volitional movement termination response. Both age groups showed similar step length, step width and swing time in the execution of the response. However, since these measures are all prior to foot-contact, these findings further support the importance of the restabilisation phase after foot-contact in regaining stability especially in the M/L direction.

With huge emphasis on the restabilisation phase as an opportunity to restore dynamic balance, Singer et al. (2014) took a closer look at the restabilisation phase and examined if there were age-related differences in kinetic indices describing the execution of the volitional step within the restabilisation phase. In this study, the researchers introduced a measure called the angle of divergence, which quantifies the extent to which the GRF_{net} act eccentrically relative to the whole-body CoM, in order to determine whether a stabilizing moment or a destabilizing moment was present. The researchers also found two distinct points (P1 and P2) throughout the restabilisation phase where the largest moments occurred. Just like in the previous study, both age groups displayed similar step lengths, step widths and swing times. Older adults, however, were found to have challenges generating appropriate eccentricities of the ground reaction force (i.e. decreased mediolateral angle of divergence compared to the

YA as well as delays in the timing of the P2 eccentricity, along with a high level of variability between trials). These findings depicted the struggles that OA have in generating the proper magnitude, timing and consistency in the orientation of the GRF to produce an adequate external stabilizing moment about the whole-body CoM to restore M/L stability.

Singer et al. (2016) continued to investigate the restabilisation phase closely in the control of M/L stability but this time during compensatory stepping rather than volitional movement termination. In this study, the authors utilized a lean-release protocol to elicit a compensatory stepping response and focused again on whole body CoM dynamics during the restabilisation phase. It was in this experiment where the researchers further defined the characteristics of P1 and P2 in the restabilisation phase, where the P1 GRF eccentricity was thought to arise due to limb stiffness on foot contact, and P2 GRF eccentricity arose in response to instability during restabilisation. Similar to Singer et al. (2011), CoM incongruity was used as the measure of stability for this study except these measures were in the frontal plane as its focus was on M/L stability. In this study, older adults were found to have an increased CoM M/L incongruity value as well as an increase in CoM M/L incongruity variability. Similar to the previous (Singer et al. 2014) study, it is also noted that the P2 timing was found to be delayed among older, relative to younger, adults. The P1 magnitude and timing as well as the P2 magnitude were comparable between the younger and older adults in this experiment, which suggested that older adults may not exhibit challenges with developing the extent of lower limb force required to restabilize in the M/L direction. It was observed that both age groups improved (reduced CoM incongruity) with trial repetition and that the delays in P2 timing among older adults could potentially be due to age-related impairments in detecting instability during restabilisation, or in generating force output rapidly to re-orient the net ground reaction force. All in all, the researchers concluded that older adults struggle in regulating M/L stability due to a poorly timed P2 peak compared to younger adults.

The last study (Singer et al., 2019) in this series explored the anterior stability of older adults during compensatory stepping. The authors used the angle of divergence as the measure of stability and focused on the restabilisation phase after foot-contact. In the A/P direction, both older and younger adults were found to have similar angle of divergence values at P2 however the differences were found in P1 instead. Older adults displayed a lower value at P1 compared to the younger adults. This depicts older adults having difficulty generating an adequate vertical GRF component at P1 which translates to a reduced angle of divergence value at P1. Since P1 is mostly influenced by preparatory muscle activation prior and limb compliance prior to step-contact (Singer et al., 2014), A/P stability using the compensatory stepping response may be regulated proactively prior to foot-contact unlike M/L stability which appears to rely on reactive control at P2. It remains unknown, however, how the individual joints of the lower limb are used to regulate stability during restabilisation, and the extent to which age-related differences exist in such parameters.

Gait Studies

As the study of the compensatory step is still ongoing, it can be challenging to find literature on certain parameters of interest. Up to this date, there is no known literature that investigates the intersegmental moment and power profiles in the compensatory stepping at the ankle, knee and hip joints. Even though this information is not available, meaningful hypotheses can still be formulated regarding the compensatory step using gait analysis data. The compensatory step has been often characterized to be very similar to the termination of gait. For starters, both gait termination and the compensatory step have the same function of stopping the body's movement. As gait has been studied more extensively compared to the compensatory step, moment magnitudes and power profiles have been investigated in gait studies. A study by Kerrigan et al. (1998) investigated the moment and power magnitudes of the ankle, knee and hip during slow walking versus fast walking in healthy older adults. In this study, it was observed that the ankle plantarflexor moment and power magnitude decreased in OA compared to YA. In addition, it was also observed that there is a decrease in the hip extension moment which also corresponded with a decrease in knee extensor activity. Some of these findings were later supported by a study conducted by DeVita & Hortobagyi (2000) who observed a torque and power redistribution in older adults during gait. In this study, older adults were observed to have less ankle plantarflexor and less knee extensor activation in gait. This resulted in a decreased plantarflexor and knee extensor moments, however, the decrease in these values were compensated by the increase in the hip extensor torque. A 50% reduction in angular impulse as well as a 39% reduction in work of the knee extensors were also observed in older adults. Both studies (Kerrigan el a., 1998 and DeVita & Hortobagyi, 2000) reveal a reduced moment output at the knee extensor muscles in older adults. Despite these similarities in findings, it is noted that the increase in hip extensor torque found in the study by DeVita & Hortobagyi (2000) contradicted the hip moment findings in the study by Kerrigan et al. (1998).

Another study that focuses on moment and power magnitudes in gait is a study by Silder et al. (2008). In this study, active and passive contributions to joint kinetics were observed in older adults during gait. It was found that there was an increase in hip power generation which the authors believe could be attributed to compensate for the weakened distal segments as older adults are found to demonstrate an increased reliance on proximal muscles rather than distal muscles for power generation (Winter et al., 1990, Kerrigan et al., 1998m DeVita & Hortobagyi, 2000). The study also found that there is a decrease in ankle plantarflexor moment with an increase in hip extensor moment. An increase in both hip moment and power magnitudes were observed in this study.

Moving Forward

Based on the literature, OA show challenges in recovering dynamic stability using the compensatory stepping response compared to the YA. These difficulties have been identified to be present in the restabilisation phase, specifically after foot-contact and occurs in both the A/P direction and M/L direction. The present study aims to utilize 3D kinematic analysis in conjunction with inverse dynamics to investigate the intersegmental (joint) moments of force in both the sagittal and frontal planes, as potential sources of the aforementioned age-related differences in stability control during compensatory stepping. Since studies have shown that older and younger adults perform the early phases of the compensatory step with high degree of similarity, the focus will be directed to the restabilisation phase where differences in timing and magnitude have been identified based on recent studies. The present study will analyze three measures—peak moment magnitude and timing of the intersegmental moments, along with the peak moment power. These measures will be examined at the ankle, knee and hip to explore joint-level differences in the execution of the compensatory step between age-groups. Peak magnitude and timings of the moments of force at each joint will be measured to provide insight on whether muscular force output, or the timing of force output, at a specific joint may be potential mechanisms that underlie previously documented age-related instability during the restabilisation phase of compensatory stepping. Further, moment powers will be measured to provide further understanding about whether such previous accounts of age-related differences in kinematic stability (CoM incongruity) may be related to the extent to which older adults are able to absorb energy to control CoM kinematics (i.e. decelerate the CoM) during retabilisation. By identifying these errors/differences, solutions to these issues may be formulated to improve balance and stability in older adults when executing the compensatory stepping response which in turn, could potentially reduce the prevalence of falls in older adults.

OBJECTIVE

The objective of this study is to investigate the A/P and M/L peak joint moment magnitude, timing of the peak moments and peak power values at the ankle, knee, and hip joints during the restabilisation phase, after foot-contact, in forward compensatory stepping. The obtained values will then be compared between the YA and the OA to identify any age-related differences amongst the groups of participants, which may be indicative of mechanisms underlying stability control challenges among older adults.

HYPOTHESES

Our hypotheses have been formulated based on the literature, for each outcome variable and for each joint in both the A/P and M/L direction. In addition, the hypotheses are written for OA in comparison to YA.

Moments and Powers in the A/P direction

In the A/P direction, the OA peak joint moment magnitude of the ankle plantarflexors are hypothesized to be lower relative to YA values based on documented ankle plantarflexion moment deficits noted in gait studies (Kerrigan et al., 1998 and DeVita & Hortobagyi, 2000) and due to the reduction in the P1 peak eccentricity found in older adults, which may have arisen from a reduced ankle plantarflexor moment. At the knee in the A/P direction, we hypothesize that the OA will also display a lower peak joint moment magnitude of the extensor muscles as gaits studies reveal reduced torque at the knee and a 50% reduction in angular impulse as well as a 39% reduction in work of the knee extensors. At the hip in the A/P direction, we hypothesize that the OA peak joint moment magnitude of the hip extensors will be higher compared to YA—based on gait studies which found that OA compensate at the hip joint for the decrease in ankle and knee moments produced.

In terms of peak power values in the A/P direction, we hypothesize the OA to have lower peak absorption (negative) power values at both the ankle plantar flexors and the knee extensors due to the reduction in peak moment magnitudes as found in gait studies. At the hip, however, we hypothesize the OA to have higher peak absorption power values of the hip extensors due to the tendency of OA to increase the hip moment as a means of compensating for the reduced distal joints.

Moments and Powers in the M/L direction

In the M/L direction, the peak ankle inversion joint moment and the peak knee abductor joint moments are hypothesized to have similar values in comparison to the younger adults due to the minimal involvement of the ankle and knee joints in the M/L direction due to their structural limitations. At the hip joint, however, we hypothesize that there will be a lower value in peak abductor joint moment magnitude. Despite the knee joint having structural limitations in the M/L direction, the muscles that are responsible for knee moments do cross both the hip and the knee joints. Even though these muscles overlap two joints, the major moment contributor in the M/L direction at the knee joint are the bones and ligaments rather than the muscles due to its anatomical limitations. The hypothesis of lower value in peak abductor joint moment magnitude in the M/L direction at the hip is due to the OA's difficulty in controlling lateral perturbations as reported in the literature.

As far as peak power values in the M/L direction go, we hypothesize that, similar to peak joint moment values, the ankle and the knee will have similar peak absorption values due to their limited involvement in this plane of movement. At the hip joint though, we hypothesize the OA to have lower peak absorption power values due to the decrease in the hypothesized peak abductor moment magnitude found in OA.

Timing

In the A/P Direction

In terms of the timing, we hypothesize that the timing of the peak moment at the ankle of older adults will be similar to the younger adults however we hypothesize that the sagittal plane peak knee and peak hip moments will be delayed. This hypothesis is based on the previous finding of a reduced P1 peak GRF_{net} eccentricity among OA, which is believed to arise from increased limb compliance on stepping limb contact, perhaps resulting from a delay in knee and hip moment generation.

In the M/L Direction

For the frontal plane, it is believed that there will be a delay in the peak hip abductor moment and extensor power. This hypothesis is formulated due to the finding of the P2 peak GRF_{net} eccentricity being delayed in OA. As the P2 GRF_{net} eccentricity is believed to arise in response to M/L instability during restabilisation, and hip joint moments are considered to be dominant in M/L stability control, a delay in hip moment generation could lead to previously observed instability among OA.

METHODOLOGY

Study Design

A cross-sectional observational study design was used in this experiment. This structure allowed us to effectively investigate the effects of age on how the compensatory stepping mechanism is executed. The participants were divided into YA and OA; both groups were asked to complete the same series of tasks and their movements/responses were analyzed to identify any age-related differences in individual joint movements of the lower limb. This study followed a cross sectional design—the participants in this study were compared only at one point in time, as they were only asked to perform the tasks on a given day without any follow ups in the future.

Participants

This study required the inclusion of a minimum of 28 participants (14 per group), which was obtained using α = 0.05, β = 0.2, f = 0.28. The effect size, f, was obtained from previous work examining age related differences in reactive control of the ground reaction force during perturbation-evoked stepping (Singer et al., 2016).

This study involved a total of forty (40) participants that were categorized in either the YA group (18-30 years old) or the OA group (65+ years old). Each group was composed of twenty (20) individuals who met the inclusion/exclusion criteria of the study, as outline below. Among the twenty individuals in each group, ten (10) were male and ten (10) were female participants.

The ages of the participants in the YA group were 24±5 years old and had a body mass of 68.1±10.3 kg while the participants in the OA group were 71±5 years old and had a body mass of 77.0±16.8 kg.

Inclusion/Exclusion Criteria

Participants had to be free from injury and must have had no self-reported neurological or cognitive impairments. The participants were able to stand or walk without the use of walking aids such as walker, canes, hiking poles, etcetera. The participants were not using psychoactive medication during the duration of the study. The participants had no reported history of previous falls that resulted into injury while performing activities of daily living. The participants must not have had any lower limb surgeries performed nor requiring any upcoming lower limb surgeries.

Instrumentation

The laboratory equipment used to conduct this experiment were the following:

1.) Eight (8) Vicon MX-3+ cameras from Vicon Motion Systems, Los Angeles, California, USA to record kinematic data at 100Hz
2.) Four (4) force plates arranged in a rectangular manner from Advanced Medical Technology Inc., Watertown, Massachusetts, USA to record kinetic data at 2000Hz
3.) Reflective markers that were 1cm in diameter from B&L Engineering, Santa Ana, California, USA

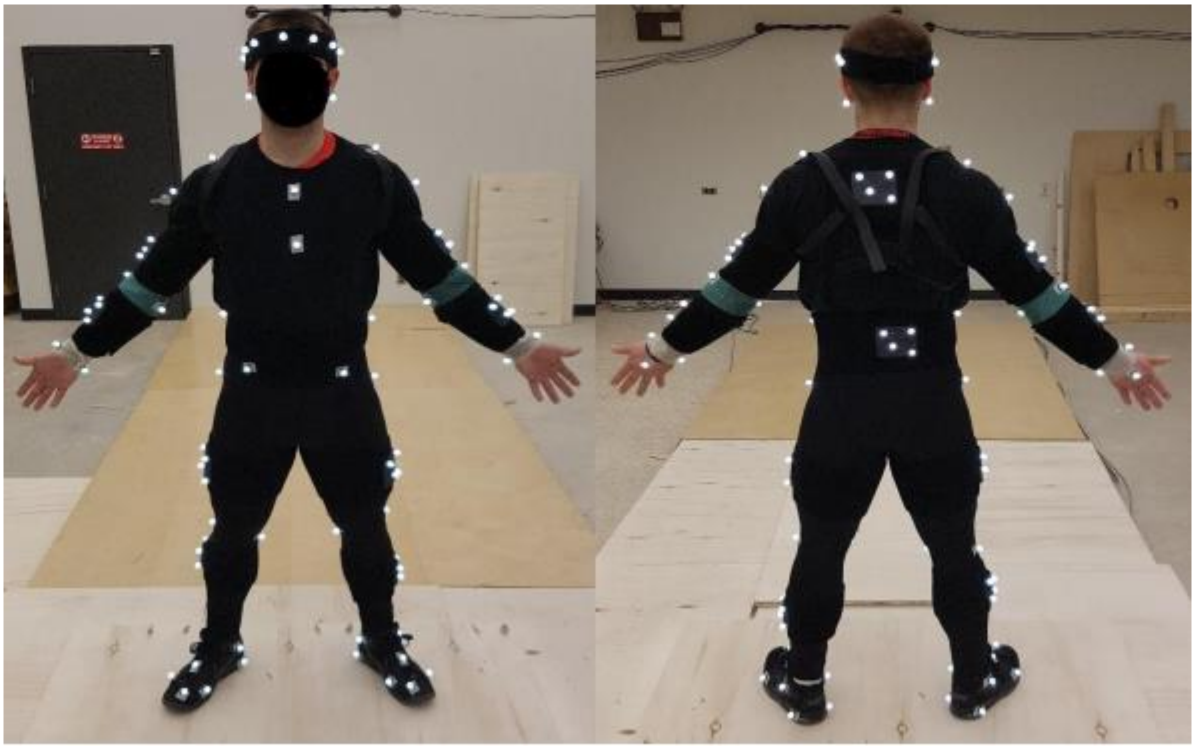
4.) Electromyogram (EMG) with the use of disposable self-adhesive Ag/AgCl electrodes from Bortec Biomedical, Calgary, Alberta, Canada to record electromyographic activity at 2000Hz.

5.) Nexus software from Vicon Motion Systems, Los Angeles, California, USA

Procedure

Clusters of four (4) reflective markers (technical clusters) were placed on each segment of the body of the participant in a similar pattern as described by Rawal et al (2021). Additional calibration markers were utilized to define segment endpoints for the head, upper limbs, trunk and pelvis. These calibration markers were placed bilaterally on the anterior aspect of the external auditory meatus, acromioclavicular joints, greater and lesser tubercles of the humerus, medial and lateral epicondyle of the elbow, radial and ulnar styloid processes of the wrist, the head of the third metacarpal, iliac crests, anterior superior iliac spine of the pelvis and the posterior superior iliac spine of the pelvis. In addition, rigid clusters of four markers were placed bilaterally on the trunk, sacrum, thighs and feet to determine 3D kinematics of each segment. Figure 2 shows the placement of the markers and rigid clusters. The electromyogram electrodes were placed bilaterally on tibialis anterior and soleus of the participant. These electrode readings were only used to examine preparatory muscle activity prior to the perturbation, which needed to be standardized between trials and between participants.

Figure 2: Placement of Markers and Rigid Clusters

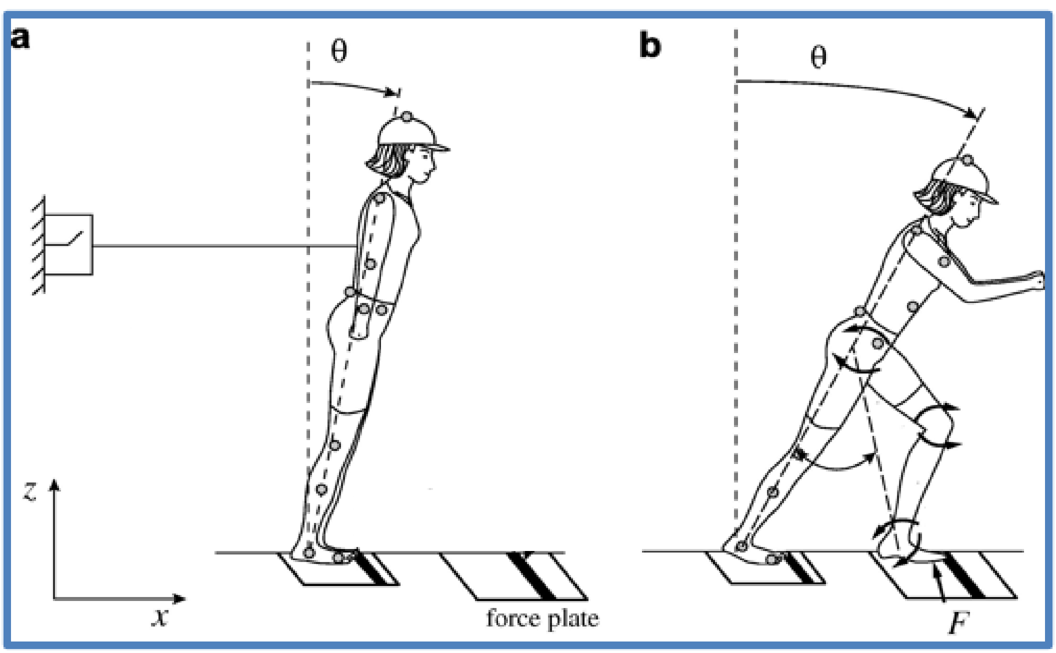


Once the participant had all markers affixed, they were asked to stand in a neutral position while a reference trial is obtained. The obtained data from the reference trial, in conjunction with anatomical calibration markers, determined the segment endpoints, the segment embedded local coordinate systems, and the transformation matrices between the local and global coordinate systems for each segment. After the reference trial was done, the markers solely used for calibration purposes were removed from the participant, leaving only the technical clusters on each segment, which tracked the position and orientation of each segment in space.

Adhesive tape was placed on the force platforms for participants to use as a guide for their starting position, such that the feet were 0.17m apart (at the heels), each with 14^{o} of external rotation relative to the forward (y) axis (McIlroy and Maki, 1997). A single quiet standing trial was collected, in which the participant was asked to stand with their feet side-by-side on separate force platforms, according to foot placement markers, for 60 seconds. This trial was used to provide a reference for the A/P CoP position and baseline EMG activity for the pre-perturbation interval.

After the quiet standing trials, the participants were attached to the lean and release mechanism (refer to Figure 3) with an initial forward lean to result in 10% of the participant's body weight being applied to the cable. Each participant performed ten (10) experimental trials. A lean of 10% of the participant's body weight was used because it is the percentage of lean that was enough to evoke a stepping response in both younger and older adults, without eliciting a multiple stepping response among the older adults

Figure 3: Lean and Release Mechanism (Hsio-Wecksler & Robinovitch, 2007)



Prior to the trial perturbation, the participant's vertical forces under each foot were monitored to ensure symmetrical weight distribution. The participant's A/P CoP position was also monitored so that it did not exceed what was observed during the initial quiet standing condition with feet side-by-side. The electromyographic activity of the tibialis anterior and the soleus muscles were also monitored to ensure they did not exceed what was observed during the initial quiet standing trial. These conditions needed to be met before the perturbation occurs. If additional muscle activity was present prior to the perturbation, the participant was provided verbal cues to relax the gastrocnemius/soleus (i.e. "allow the cable to hold you up/try not to use your ankle muscles to hold you upright"), which helped achieve these specific conditions. Once the participant met the starting conditions, a tone was played which signals the beginning of the trial. To limit the effects of anticipation, the cable release onset was performed at different time intervals. Participants were asked to recover from the perturbation using a preferred step length, width and speed, and remain in the final stable position for ten (10) seconds. In addition, five (5) catch trials were randomly presented to the participants. In these trials, there was no elicited perturbation and participants remained in the forward leaning position. These catch trials were included to prevent the participants from becoming accustomed to the perturbation magnitude involved in the experiment. Analysis of data was only applied to successful single-step balance recovery responses by the participant.

Analysis

The outcome variables used in this study were: peak lower limb intersegmental moment of force magnitude—normalized to body mass (N•m/kg), time to peak moment(s), peak moment power—normalized to body mass (W/kg), for each of the three joints of the lower limb, and time to peak power. These variables were measured in both the anterior/posterior (A/P) direction and the medial/lateral (M/L) direction. The acquired measures were then compared between younger adults versus older adults to identify any age-related differences in the execution of the restabilisation phase of the compensatory step. This may help elucidate the specific joint(s) responsible for previous literature findings of instability among OA during the restabilisation phase following sumbaximal-anterior balance perturbation.

Force plate data were lowpass filtered with a zero-lag, 4th order, Butterworth filter with a cutoff frequency of 50Hz. Marker data were lowpass filtered with zero-lag, 20th order critically damped filter with a cutoff frequency of 6Hz (Robertson and Dowling, 2003). Filters applied to both kinematic and kinetic data reduce high frequency noise caused by errors in automatic processing of centroid locations as well as errors caused by vibrations recorded by the force platforms.

Modelling of the entire body was done as a rigid system of independently tracked segments. For older adults, Dempster's segment parameters (Dempster, 1955) was used to estimate segment masses. For centre of mass positions, these were estimated based on the geometrical model used by Hanavan (1964) whereby the head was modelled as a sphere, the pelvis and trunk were modelled as cylinders and the appendicular skeleton was modelled as conical frusta (i.e. an elongated truncated cone). For younger adults, Dempster's segment parameters were not representative (based on the ages of cadavers included in the study (Dempster, 1955)) so the segment masses and centres of mass positions were estimated using the values reported by Zatsiorsky et al. (1990) which were later modified by de Leva (1996).

Data analysis primarily involved the use of inverse dynamics to calculate the joint moment magnitudes of each joint of the lower limb. Joint moment magnitude was obtained through the use of the Newton-Euler equations. Joint power was obtained as the product of joint moment magnitude and joint angular velocity. Although power is a scalar quantity, it was calculated within each plane by the scalar multiplication of the intersegmental moment and angular velocity (Eng and Winter, 1995). For joint moment magnitude values, a positive moment in the sagittal plane at the ankle is dorsiflexor, at the knee is extensor and at the hip is flexor. A positive moment in the frontal plane for the right ankle is invertor, for the right knee is adductor and for the right hip is adductor. As positive rotations about a forward-directed anteroposterior axis are counterclockwise, a positive rotation for the left limb would yield the opposite anatomical interpretation for the left limb in the frontal plane (i.e. ankle evertor, knee abductor, hip abductor). For the transverse plane, a positive moment would be internal rotation for all joints on the right side and would be for opposite (external rotation) for the left side joints. In terms of joint power values, regardless of plane or axis of rotation, a positive joint power value represents a concentric muscle contraction (energy generation) while a negative joint power value represents an eccentric muscle contraction (energy absorption).

Each outcome variable was subjected to an independent samples t-test ($p<0.05$) to identify differences between age-groups in each outcome variable. The Shapiro-Wilk test will be used to evaluate the normality assumption. Levene's test will be used to evaluate the homogeneity of variance assumption. Should a variable violate this assumption, the degrees of freedom will be adjusted using the Welsh-Satterthwaite method—the adjusted degrees of freedom will be used to determine the critical t-value at $p<0.05$. There are three outcome variables (peak joint moment magnitude, timing of the peak moment and the peak joint power) and two planes of axes (sagittal and frontal plane) that are involved which accounts for six T-tests for each of the three joints involved.

Secondary analyses were also performed to ensure group differences that emerged during the restabilisation phase were not as a result of differences that existed prior to foot contact. Initial variables analyses were utilized in order to properly execute the methods of this study to establish similarity in the initial conditions between participants and trials—CoM displacement following the APA and during the APA (in both the sagittal plane and the frontal plane) was measured to depict the effect of the APA while the CoM velocity at step contact (in both the sagittal plane and frontal plane) was measured to depict the similarities in perturbation. These initial variables are the variables that have been measured prior to the restabilisation phase in the compensatory stepping mechanism. In addition, statistical analysis of the margin of stability (MoS) was also performed to evaluate the level of stability of each participant at certain time points: at instant of perturbation onset, at foot-off (beginning of swing phase) and at foot-contact (end of swing phase and the start of the restabilisation phase). MoS values were measured in meters and were used in this experiment to quantify the participant’s level of stability. MoS depicts the relationship between the extrapolated CoM position (XCoM) and the BoS at a given time—a positive MoS value indicates that the CoM will be within the BoS while a negative MoS value indicates that the XCoM will be outside of the body's BoS at a certain instant in time. The MoS was also examined during the restabilisation phase a certain time points: at its minimum, the time from foot contact to the minimum MoS, and the change in MoS from the onset of restabilisation (MoS at foot-contact) to the minimum MoS during restabilisation.

RESULTS

Data were collected from forty (40) participants—twenty (20) OA and twenty (20) YA, each group being composed of 50% female. All participants included met the criteria for inclusion/exclusion as outlined in the methods. The primary results of the experiment along with secondary analyses are reported below. The p value used in the study was p = 0.05.

Primary Analyses

Peak Moment

Older adults exhibited a significantly reduced ankle plantarflexor moment compared to YA ($t(26.66) = -2.073$, p=0.048, d=0.140). There were no other significant age-related differences in peak intersegmental moments. Table 1 contains peak moment results.

Table 1: Peak Intersegmental Moments Among Older Adults and Younger Adults

		Older Adults (n=20)		Younger Adults (n=20)	
Plane	Moment (Nm/kg)	Mean	Std Deviation	Mean	Std Deviation
Sagittal	Ankle Plantarflexor*	0.6482	0.0824	0.7398	0.1795
	Knee Extensor	0.5231	0.3060	0.5192	0.2499
	Hip Extensor	1.3190	0.2640	1.4201	0.4001
Frontal	Ankle Evertor	0.0283	0.0299	0.0155	0.0184
	Knee Abductor	0.1769	0.1130	0.2015	0.1088
	Hip Abductor	0.4579	0.1487	0.4507	0.1417

* $p < 0.05$

Peak Power

OA produced a significantly reduced peak ankle absorption power in both the sagittal ($t(38) = -2.895$, p=0.006, d=0.155) and the frontal plane ($t(38) = -2.640$, p=0.012, d=0.039) compared to the YA. Table 2 contains power results.

Table 2: Peak Eccentric (Absorption) Intersegmental Moment-Powers between Older Adults and Younger Adults

		Older Adults (*n*=20)		Younger Adults (*n*=20)	
Plane	Eccentric Power (W/kg)	Mean	Std Deviation	Mean	Std Deviation
Sagittal	Ankle*	-0.3487	0.1345	-0.4903	0.1724
	Knee	-0.9056	0.5298	-0.7423	0.4176
	Hip	-0.7821	0.9325	-0.6911	0.6261
Frontal	Ankle*	-0.0453	0.0378	-0.0776	0.0394
	Knee	-0.0687	0.0548	-0.0490	0.0544
	Hip	-0.2853	0.2894	-0.2053	0.0894

* $p < 0.05$

Time to Peak Moment

OA exhibited a longer time to peak hip extensor moment (sagittal plane) (t(19.889) = -2.222, p=0.038, d=0.063) however, OA exhibited shorter time to peak knee abductor moment (frontal plane) (t(38) = 3.397, p=0.002, d=0.172). Table 3 contains time to peak moment results.

Table 3: Time to Peak Intersegmental Moment Results between Older Adults and Younger Adults

		Older Adults (n=20)		Younger Adults (n=20)	
Plane	Time to Peak Moment (s)	Mean	Std Deviation	Mean	Std Deviation
Sagittal	Ankle	0.5727	0.4200	0.5082	0.4348
	Knee	0.1835	0.1284	0.1468	0.0832
	Hip*	0.0999	0.0880	0.0557	0.0135
Frontal	Ankle	1.1160	1.9177	0.7817	1.5322
	Knee**	0.1002	0.1009	0.2853	0.2218
	Hip	0.1464	0.0792	0.1746	0.1083

* $p < 0.05$
** $p < 0.01$

Time to Peak Power

In the sagittal plane, OA exhibited an increased time to peak eccentric power at the ankle (t(21.046) = -2.634, p=0.015, d=0.0768) and hip (t(28.513) = -2.441, p=0.021, d=0.299), along with a reduced time to peak eccentric power at the knee (t(38) = 2.610, p=0.013, d=0.059) in relation to YA.

Similarly, in the frontal plane, OA again exhibited an increased time to peak eccentric power at the ankle (t(20.980) = -2.672, p=0.014, d=0.201) and hip (t(24.275) = -2.967, p=0.007,d=0.061), however there was no significant reduction in time to peak eccentric power at the knee in relation to YA, as would be predicted given the earlier peak knee abductor moment, previously noted. Table 4 contains results for time to peak power.

Table 4: Time to Peak Eccentric (Absorption) Power Results between Older Adults and Younger Adults

		Older Adults (n=20)		Younger Adults (n=20)	
Plane	Time to Peak Eccentric Power (s)	Mean	Std Deviation	Mean	Std Deviation
Sagittal	Ankle*	0.1761	0.1058	0.1121	0.0246
	Knee**	0.0501	0.0603	0.0989	0.0579
	Hip*	0.2442	0.2348	0.0981	0.1039
Frontal	Ankle*	0.2675	0.2765	0.0981	0.0632
	Knee	0.1376	0.1050	0.1546	0.1317
	Hip *	0.1764	0.0803	0.1194	0.0302

* $p < 0.05$
** $p < 0.01$

Secondary Analyses

Initial Variables

There were no significant age-related differences in any variable measured prior to the onset of the restabilisation phase (Table 5).

Table 5: Initial Movement Variables Measured Prior to the Onset of the Restabilisation Phase

	Older Adults (*n*=20)		Younger Adults (*n*=20)	
Variable	Mean	Std Deviation	Mean	Std Deviation
CoM Displacement following the APA (m)	0.0048	0.0040	0.0059	0.0050
CoP Displacement during the APA (m)	0.0335	0.0177	0.0468	0.0256
M/L CoM Velocity at Step Contact (m/s)	0.1724	0.0478	0.1811	0.0576
A/P CoM Velocity at Step Contact (m/s)	0.5216	0.0869	0.5505	0.1130
Step Width (m)	0.2888	0.0283	0.2830	0.0274
Step Length (m)	0.4462	0.0739	0.4462	0.0805

Margin of Stability

Margin of stability (MoS) values were obtained both prior to the restabilisation phase and during the restabilisation phase.

Table 6: Margin of Stability Before the Restabilisation Phase between Older Adults and Younger Adults

		Older Adults (*n*=20)		Younger Adults (*n*=20)	
Plane	Margin of Stability (m)	Mean	Std Deviation	Mean	Std Deviation
Sagittal	MoS at perturbation onset	0.0518	0.0279	0.0501	0.0248
	MoS at foot-off	-0.0756	0.0425	-0.0830	0.0355
	MoS at foot-contact	0.1840	0.0393	0.1735	0.0317
Frontal	MoS at perturbation onset	0.1738	0.0128	0.1733	0.0084
	MoS at foot-off	0.1653	0.0249	0.1567	0.0218
	MoS at foot-contact*	0.1344	0.0193	0.1168	0.0184

Note: A positive MoS indicates an extrapolated CoM (XCoM) positioned within the BoS; a negative MoS indicates an XCoM outside the BoS
* $p < 0.05$

It was found that the OA exhibited greater frontal plane MoS at the instant of foot contact (signifying great stability) than the YA ($t(38) = -2.939$, p=0.006,d=0.019) (Table 6). While there were no significant differences in the MoS at foot-contact in the M/L direction, OA exhibited a larger change in the MoS between foot-contact at the point at which the MoS reached its minimum value ($t(38) = -2.686$, p=0.011,d=0.007) (Table 7).

Table 7: Margin of Stability Values during Restabilisation Phase between Older Adults and Younger Adults

		Older Adults (n=20)		Younger Adults (n=20)	
Plane	Variable	Mean	Std Deviation	Mean	Std Deviation
Sagittal	Min MoS (m)	0.1820	0.0395	0.1704	0.0319
	Time to Min. MoS (s)	0.0051	0.0228	0.0000	0.0000
	MoS Change (m)	0.0020	0.0042	0.0031	0.0064
Frontal	Min MoS (m)	0.1176	0.0188	0.1059	0.0195
	Time to Min. MoS (s)	0.1008	0.0734	0.1005	0.1897
	MoS Change (m)*	0.0167	0.0072	0.0109	0.0065

* $p < 0.05$

Summary of Hypotheses and Results

Due to the vast amount of variables involved in this study, Table 8 provides a summary of results to make it easier to track and compare our formulated hypotheses, the results of our study and whether or not the results agree with our hypotheses.

Table 8
Summary of results versus hypotheses made in the experiment

Joint	Plane	Outcome Measure	Hypothesis	Results	Agree/Disagree
Ankle	Sagittal	Peak Moment*	↓	↓	yes
		Peak Moment Time	Ø	Ø	yes
		Peak Power*	↓	↓	yes
		Peak Power Time*	Ø	↑	no
	Frontal	Peak Moment	Ø	Ø	yes
		Peak Moment Time	Ø	Ø	yes
		Peak Power*	Ø	↓	no
		Peak Power Time*	Ø	↑	no
Knee	Sagittal	Peak Moment	↓	Ø	no
		Peak Moment Time	↑	Ø	no
		Peak Power	↓	Ø	no
		Peak Power Time*	↑	↓	no
	Frontal	Peak Moment	Ø	Ø	yes
		Peak Moment Time*	↑	↓	no
		Peak Power	Ø	Ø	yes
		Peak Power Time	↑	Ø	no
Hip	Sagittal	Peak Moment	↑	Ø	no
		Peak Moment Time*	↑	↑	yes
		Peak Power	↑	Ø	no
		Peak Power Time*	↑	↑	yes
	Frontal	Peak Moment	↓	Ø	no
		Peak Moment Time	↑	Ø	no
		Peak Power	↓	Ø	no
		Peak Power Time*	↑	↑	yes

Note:
↑ means that the OA will exhibit significant higher values relative to YA
↓ means that the OA will exhibit significant lower values relative to YA
Ø means that there are no significant differences between OA and YA
* means that there is a joint specific difference between OA and YA

DISCUSSION

Given our previous knowledge of age-related differences in whole-body stability control, the goal of this study was to determine if age-related differences in ankle, knee and hip kinetics are present during the execution of a single compensatory step, and may be responsible for previous reports of age-related instability (McIlory et al., 1996, Singer et al., 2016, 2019). Such knowledge moves from knowledge of whole-body, global, stability control mechanisms (Horak & McPherson, 1996, McPherson, 1991 and Lee, 1984), to local, joint level, factors that could present targets for balance training. The present work focuses on the peak joint moments, peak joint powers and the timing of these variables in both the sagittal and frontal planes. The results exhibit a mixed agreement with our formulated hypotheses for this study. The following discussion will aim to shed light on the kinetic sources of age-related differences in stability and also provide information about potential mechanisms of age-related instability.

Whole Body Differences

Initial Movement Parameters

In order to identify age-related differences in the restabilisation phase of the compensatory step, it is imperative to examine the initial movement parameters, which occur prior to stepping limb contact, to ensure that the observed differences during restabilisation are not the result of lingering effects that emerged in prior phases of the movement. Measures of the anticipatory postural adjustment (APA), the effect of the APA on CoM displacement, CoM velocity, step width/length, along with the MoS at perturbation onset and at foot lift-off exhibited no significant differences between groups

(Tables 5 and 6). This between group similarity in the MoS at perturbation onset suggests that, on average, the perturbations that both the OA and the YA experienced were similar in magnitude. In addition, while an APA was present in both groups, the APA had minimal effect, as evidenced by both the lateral CoM displacement following the APA (Table 5) and the MoS at foot-off (Table 6). The lack of significant differences in these initial movement parameters, while not indicative of equivalence, suggests that the lean-and-release protocol utilized in this study was effective at mitigating the effects of any meaningful anticipatory movement strategies that YA and OA may apply to prepare for the upcoming perturbation. As the OA group was not more unstable at the onset of the restabilisation phase, examination of group differences during the restabilisation phase can occur largely unencumbered by the necessity to factor out the influence of prior instability on restabilisation phase dynamics.

Nevertheless, it is important to bear in mind that the cumulative effects of many non-significant age-related differences in initial movement parameters can be influential on restabilisation phase dynamics. This may be the case with the larger frontal plane MoS at the instant of foot-contact among the OA group, suggesting greater ML stability among OA at the onset of the restabilisation phase. This may result from a non-significant age-related reduction in ML CoM velocity at step contact coupled with an increase in step width (Table 5), as both variables are included in the calculation of the MoS. From this, it appears that the OA group began the restabilisation phase in a more stable state than the YA group and, as such, it will be necessary to contextualize age-related differences in restabilisation phase dynamics with this in mind.

Age Related Differences in Stability During Restabilisation: Margin of Stability

By examining both the frontal and sagittal plane MoS values during the restabilisation phase, we found that there were no differences in the minimum MoS and time to minimum ML MoS between OA and YA. This means that the OA and the YA were experiencing similar levels of instability and the time course of events leading to reversal of the COM velocity vector were similar between groups in both the sagittal and frontal planes. However, as the OA began the restabilisation phase in a more stable position than the YA (as signified by the frontal plane MoS at foot contact) but exhibited similar levels of instability during the restabilisation phase (as signified by the minimum MoS), we felt it necessary to compute the change in MoS that occurred between the onset of restabilisation and when the MoS was at its minimum value. From examining these values, differences were seen in the MoS change in the frontal plane, with the OA exhibiting a greater change in stability compared to the YA during the restabilisation phase. Despite the focus on, and importance of, restabilisation phase parameters in previous work (Singer et al., 2011, 2014, 2016), this observation points to the potential importance of the initial movement parameters and anticipatory control in mitigating instability among the OA group. As such, had the OA not made an accumulation of small (but non-significant) augmentations in ML CoM velocity and step width and, consequently, not been in a more stable position at the onset of restabilisation, it is possible that the OA would have become more unstable during restabilisation. The greater change in MoS during the restabilisation phase points to challenges among OA in either perceiving instability after foot-contact or producing optimal magnitude or direction of applied force when recovering from the perturbation. It is important to note that previous work (Singer et al., 2016) has observed improvements in kinematic measures of stability during the restabilisation phase (termed COM incongruity) with trial repetition. It is possible that there are also improvements in stability during the initial (initiation and swing) phases of the compensatory stepping response as individuals learn the dynamics of the perturbation. The present work did not examine the potential for such improvements; future work should examine the potential for stability improvement through practice-related modifications of the initial movement parameters.

Joint Specific Differences

Ankle Peak Moment and Peak Power

Aside from whole-body differences revealed by the MoS, joint-specific differences between OA and YA were also observed during the restabilisation phase, which may underlie age-related differences in stability following an external balance perturbation. In terms of peak moments in the A/P direction, the ankle results agreed with our hypothesis in which the OA would exhibit a significantly decreased peak plantarflexor moment value compared to the YA. The plantarflexor moment is one of the major contributors to balance recovery in the sagittal plane as it greatly influences the CoP (Kerrigan et al., 1998, Ema et al., 2016). When we experience a forward disruption of our balance, the role of the plantarflexor moment in the compensatory step is to shift the CoP forward and to restrain the forward rotation of the tibia over the talus (i.e. constrain ankle dorsiflexion). This allows for the effective transmission of force to the ground, generated by extensor moments at the hip and knee, which creates an appropriate magnitude and orientation of stabilizing force to stop the body from falling forward. As such, the ankle intersegmental moment may be particularly important for creating a stable platform to facilitate effective force transmission to the ground, from more proximal joints. In recovering from each perturbation, there is likely a certain threshold of net stabilizing force that must be met in order to recover from a perturbation. If the threshold is not met, this could result in an irregular compensatory stepping pattern such as a multiple step response, or even a loss of balance. While there were no age-related differences in frontal plane ankle moments, the reduction in the plantarflexor moment may also contribute to the increased change in M/L MoS among the OA group, as individuals typically land with their foot slightly externally rotated. As such, the impact of the plantarflexor moment may not be specific to sagittal plane CoM dynamics alone, but may also contribute in some extent to M/L restabilisation.

Interestingly, age-related differences in peak moment magnitude were only found at the ankle, which suggests that muscle moment production at joints of the lower limb may not be a primary limiting factor in the restabilisation response. Nevertheless, there could be a few reasons for the age-related difference at the ankle joint. One of the reasons could perhaps be due to an inability of the OA to generate the same “quantity” of plantarflexor moment compared to the YA in a short amount of time (i.e. in the context of angular impulse, which is the area under the moment-time curve). If this is the case, the peak plantarflexor moment would be reduced, but the overall amount of plantarflexor activity, revealed by the plantarflexor angular impulse, would be maintained. While this would be effective in restraining forward (and perhaps even lateral) translation of the CoM in relation to the limits of the BoS (such that the OA group did not exhibit loss of balance or multiple stepping responses), it could increase the time duration over which the COM was permitted to translate anteriorly or laterally (i.e. reduced COM deceleration). This could have led to our observation of an increased change in MoS during restabilisation, as the CoM would translate farther towards the limits of stability during this phase of the response. Another reason could be due to a perceptual challenge among OA, in which they perceive that they are exerting the appropriate amount of plantarflexor moment but is inadequate to recover from the perturbation. The question remains, however, as to why such a perceptual issue would not equally affect force/moment output from all segments/joints involved in the compensatory stepping response. As results at the ankle in the mediolateral direction were consistent with our hypothesis, in which both the OA and the YA exhibited similarity in peak moment values at the ankle joint, it is possible that challenges with motor output among older adults are specific to the ankle plantarflexors (but could also arise because of the potential limited role of the ankle invertors/evertors in restraining M/L CoM displacements). Previous works have recorded specific deficits in generating rapid ankle plantarflexor moment production among OA (Ema et al., 2016, Franz, 2016) but the question remains regarding the specific sensorimotor mechanism underlying why OA underutilize ankle plantarflexors to regain stability. As maximal ankle moment output was likely not required to restabilize from the perturbation magnitude used in this study, it seems unlikely that age-related differences in strength played a large role. It may be more likely that age-related differences arose as the OA either face challenges perceiving the appropriate moments required to recovery stability from a given perturbation, or experience difficulty perceiving the state (position/velocity) of their CoM during the restabilisation phase.

When it comes to the peak ankle powers, we observed reductions among OA in both the A/P and M/L directions, which was consistent and inconsistent with our hypothesis, respectively. Previous studies have examined the age-related differences in OA versus YA in executing the compensatory step after a perturbation but these studies were only limited to observing the peak ankle moment (King et al., 2011). With this in mind, we formulated our hypotheses that as there will be a decrease in the peak plantarflexor moment in OA, it would also result in a decrease in peak ankle power. Ankle power can be calculated by using the formula:

$$\text{Power} = \text{Moment of Force} \times \text{Angular Velocity}$$

This means that power has a direct relationship with moment of force and angular displacement but an inverse relationship with time (as angular velocity = angular displacement/time). By comparing the ankle moments in the A/P direction of the results, it is directly observed that the OA demonstrated a decrease in plantarflexor moment compared to the YA. As a result, it is expected that the peak ankle peak power will also decrease, as seen in our data, as there is reduced energy absorption in a given amount of time. In the M/L direction however, we observed a significant difference in the ankle peak power but we do not observe a difference in the ankle peak moment. This could mean that this difference is due to an increase in time to peak moment (or a reduction in joint angular velocity) with the OA compared to the YA when it pertains to ankle plantarflexion. Our data do show that there is a reduction in the peak evertor moment and an increase in time to peak ankle evertor moment in OA however neither difference is statistically significant. It is possible that the peak ankle absorption power was statistically significant as small age-related differences in joint moment and time (which would be captured in the time to peak moment and/or joint angular velocity, which was not measured) became additive when analyzing joint power. Further examining the ankle moment and time to peak moment results, there is high between-participant variability in these data. This reduced statistical power could perhaps be a reason why significant differences in this metric were not noted.

Knee Peak Moment and Peak Power

At the knee joint, the peak moment results in the A/P direction did not agree with our hypothesis. We hypothesized that the OA would exhibit a lower peak moment in the sagittal plane compared to the YA, however our results show that there is no significant difference in the peak moment values between the OA and the YA. Older adults have been shown to exhibit an increased incidence of double stepping responses (McIlroy & Maki, 1996, Thelen et al., 1997); we hypothesized that a double stepping response could represent a strategy to compensate for the decrease in knee moment generation—having a second step could be an opportunity for the OA to generate additional knee stabilizing moments in a subsequent step. It is possible that age-related differences in A/P stability do not arise from age-related differences in knee extensor moment output, but instead arise more from deficits at the ankle or hip. It may also be possible that the lack of significant differences at the knee joint could be as a result of the instructions given to participants to try to restabilize using a single step. If multiple stepping responses are a “strategy” as opposed to arising out of necessity to decelerate the CoM after an ineffective initial step, such instructions could motivate the OA to abandon the multiple step strategy and increase lower limb moment output. Future work could limit specific instructions given to participants, such that they are free to self-select a strategy to restabilize.

With regards to peak knee power values at the knee joint, our results did not agree with our hypothesis. We hypothesized that the OA would display a decreased peak knee power value compared to the YA, however based on our results, we found that there are no significant differences between the peak power values at the knee between OA and YA. One possible explanation for the lack of agreement with our hypothesis is because we based our knee power hypotheses on the assumption that the knee stabilizing moments will display a significant difference, which was not the case according to our actual results. Another explanation for the lack of differences is due to the simple fact that the knee joint power is not a critical factor in the recovery of stability using a single compensatory step, but instead is used more isometrically for energy transmission and dissipation at other joints.

Hip Peak Moment and Peak Power

Lastly, at the hip joint, our hypotheses also did not agree with the results and no significant differences were recorded between the OA and YA in both peak moment and peak power at the hip joint in both the sagittal plane and the frontal plane. Based on the reviewed literature (King et al., 2011) we expected an increased dependency on the hip joint in generating stabilizing forces when recovering from a perturbation however, this was not the case according to our study. Similar to the previous explanations, we believe that a possible explanation to this is due to the age-related differences in moment and power magnitude occurring primarily at the ankle joint. The hip joint contains very large muscle groups that are capable of generating strong stabilizing moments. By not increasing moment output at the hip joint to compensate for the reductions at the ankle joint, this could contribute to a poor prognosis for stability when recovering from a perturbation. Nevertheless, age-related differences in hip moment output do not appear to play a role in reductions in stability experiences by OA during the compensatory stepping response.

Timing Differences at the Ankle, Knee and Hip

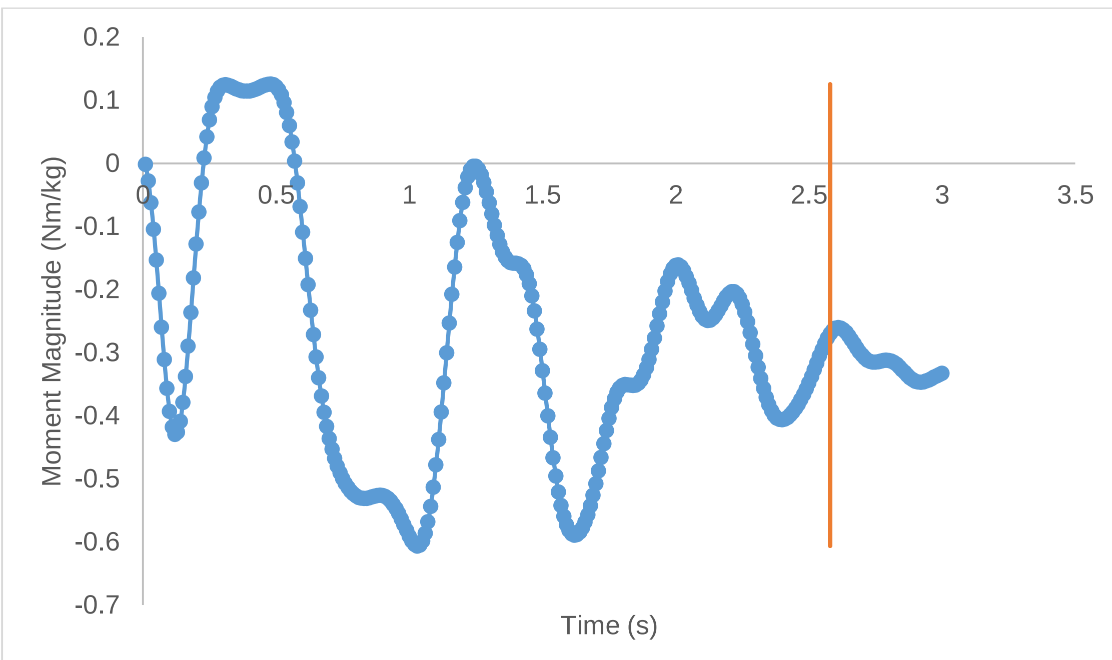
Although we only found three significant differences (ankle peak moment in the A/P direction and ankle peak power in both the A/P and M/L direction) in joint kinetics, multiple significant differences were observed within the timing of peak moments and peak powers. Before we discuss the joint specific differences in timing, it is important to understand the order of muscle activation in compensatory stepping. It is crucial to examine timing differences as a whole because changes in temporal sequencing of individual limb joint moments could potentially have an effect on the entire outcome of the execution of the single compensatory step. Our hypotheses for time to peak moment and time to peak power stated that both the OA and the YA will peak at the same time at the ankle joint but then delays occur at the knee and hip joints for OA. Our results from the experiment did not agree with our hypotheses at all joints and in both planes of movement. The reason for this major difference is due to our assumption that the stepping limb during the restabilisation phase of the compensatory step would exhibit a bottom-up muscle activation pattern consistent with the work of Tirosh and Sparrow (2005), in which the soleus, vastus lateralis and gluteus maximus were activated respectively, prior to step contact. We believed that this orderly sequence would also be present when examining the timing of the peak moments after foot-contact. This was not the case and we found that the peak moment and peak power sequence in the compensatory stepping mechanism predominantly follows the reverse order which is a top-down sequencing pattern when the hip moment and powers would peak first followed by the knee and then the ankle. Age-related differences in the sequencing of peak moment could have implications with force transmission to the ground and could hamper deceleration of the whole body CoM. Age-related differences in the timing of each variable at each joint are discussed below.

Timings of Peak Moment and Peak Power at the Ankle. Our data indicate that the OA exhibit a delay in achieving peak ankle moments relative to the YA, however the recorded differences were not noted to be statistically significant. In terms of peak power timings, the OA displayed a significant delay in compared to the YA in both the sagittal plane and the frontal plane. Such a delay in energy absorption at the ankle in both sagittal and frontal planes could have an impact on prolonging the time over which the CoM is decelerated and could lead to a larger lateral displacement of the CoM, consistent with previous studies of larger A/P and M/L incongruity magnitude among OA (Singer et al., 2016, 2019), and in the present work when considering the change in MoS from step contact to the minimum MoS. Though studies have been done measuring the peak moment capacity of OA, we have not come across joint specific studies that focus on the timing of peak moments and peak powers. Further studies could be conducted to examine this in more detail to understand the association between the timing of the peak moment and powers with instability.

Although anecdotal evidence, and not examined statistically within the present work, another possible source of delay in moments and powers could be due to the presence of an insufficient initial peak in the moment profiles among many of the older adults.

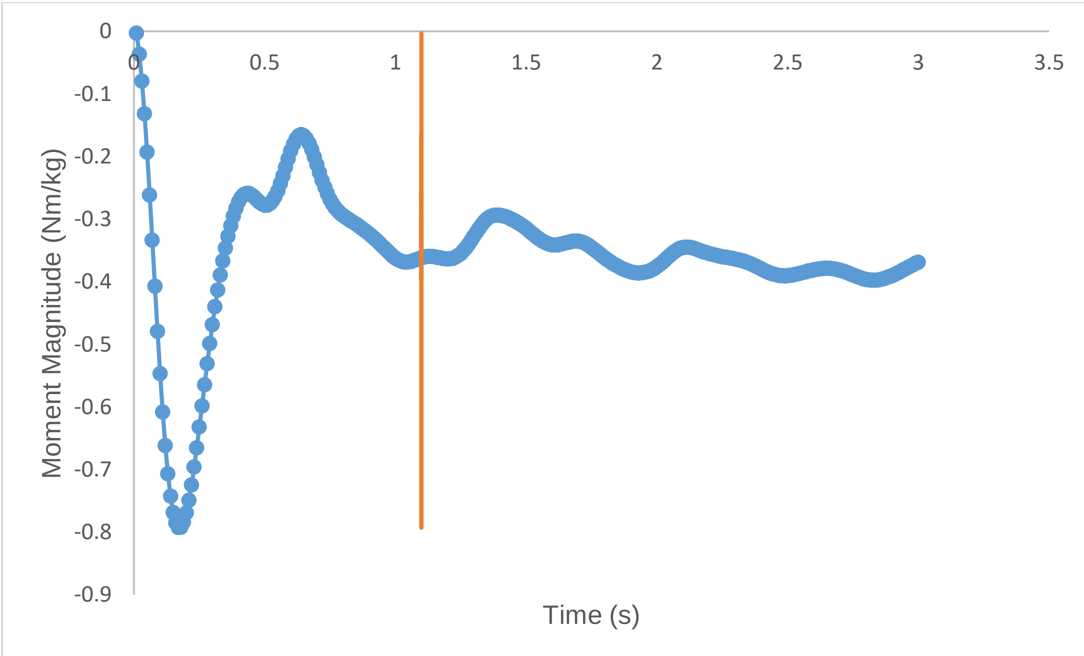
Among many trials and participants, a strong initial peak of stabilizing moment which is then followed by lower moment values was observed. Throughout our study, we noticed that the OA exhibited an insufficient initial peak, followed by a larger peak moment at every joint more frequently than the YA in the execution of the compensatory step, as is evident in the figure below (Figure 4 and Figure 5).

Figure 4: Moment Graph Depicting an Insufficient Initial Peak at the Ankle Joint in the Sagittal Plane



Note: t = 0 is foot contact; restabilisation time = 2.58s
Vertical line denotes when restabilisation occurred

Figure 5: Moment graph depicting an appropriate initial peak at the ankle joint in the sagittal plane



Note: t = 0 is foot contact; restabilisation time = 1.1s
Vertical line denotes when restabilisation occurred

An insufficient initial peak moment may have implications to the extent of instability experienced during the compensatory step, as it appears to be associated with a delay in the peak moment. In the above example, the time to peak moment without overshooting is 0.63s while the time to peak moment with the appropriate initial peak is 0.34s, which suggests there may be a significant delay created by the initial and secondary peaks. The lack of appropriately scaled initial stabilizing moments generated by OA further supports the idea that OA may make errors in accurately initially detecting either the perturbation magnitude or the level of their own joint moment output in response to the level of instability during restabilisation. As a result, there is a second attempt to generate the appropriate moment, as is seen in Figure 4. Effectively scaling the initial moment output may be a key factor in a successful recovery of balance through the compensatory step, which may be linked with the delay in reactive force output observed in prior work (Singer et al., 2014, 2016). As the OA appear to be able to generate a second larger moment, this appears to also support the idea that age-related differences in stability during submaximal compensatory stepping responses are likely not due to differences in muscular strength.

Timings of Peak Moment and Peak Power at the Knee. With respect to timings at the knee joint, the OA were found to achieve the peak frontal plane knee moment and sagittal plane knee power both earlier and first in sequence in comparison to the YA in the sagittal plane. Interestingly, these are two of the three metrics in which the OA outperform the YA in such a way where the age-related difference could be beneficial to stability. Despite this suggestion, it may be possible that the lack of top-down sequencing of peaks (i.e. in this case the knee first, followed by the hip and finally the ankle) may interfere with effective lower limb force generation and energy absorption to decelerate the CoM. While we examined age-related differences in the peak moments and powers, further work should investigate the temporal sequencing of these kinetic variables, given the potential for within-limb (across joint) coordination of force and energy absorption may be potential area of improvement that can be used in fall prevention programs in OA. Another possible explanation to the early peaking of knee moments is that these moments are generated by bone-on-bone articulation rather than muscle activity.

Timings of Peak Moment and Peak Power at the Hip. Our data indicate that the OA exhibit a statistically significant delay at time to peak moment at the hip joint in the sagittal plane compared to the YA. These results show that the mean time difference between both groups in this particular variable is nearly double which is important to note as large moments occur at the hip and delays could prolong the deceleration phase in the A/P direction and allow the CoM to travel farther toward the anterior limits of the BoS. In the frontal plane, there were no significant differences noted in the timing of peak hip moments though it is noted that the OA have a faster peak hip moment time compared to the YA. This could suggest that the OA prioritize lateral stability more than anterior/posterior stability compared to the YA as an attempt to reduce lateral falls and, potentially, hip fractures.

With respect to timing of peak powers at the hip, we observed significant differences in both the sagittal plane and the frontal plane where the OA exhibit a delayed peak relative to the YA. Again, based on the top-down sequencing of peaks, a delayed hip peak power potentially could cause delays in absorption of energy which means that there could be an increase in time that the CoM translated toward the limits of the BoS during the restabilisation phase.

CONCLUSION

Through the conducted study, age-relate differences were identified at the joint level between older adults and younger adults in the execution of the single-step forward compensatory step in recovering stability from a perturbation. When it comes to moments and powers, significant differences were found only at the ankle joint in which the older adults showed a decreased peak moment magnitude in the sagittal plane but also a decreased peak power magnitude in both the sagittal plane and the frontal plane. Despite these findings, most of the joint specific differences were actually found in the timing of peak moments and powers in the execution of the compensatory step. We found that the older adults showed significant delays in peak hip joint moments but they were found to outperform the younger adults in reaching peak moments at the knee in the frontal plane. Through this, we have concluded that the older adults may utilize a different strategy in the execution of the compensatory step. Similarly, timings of peak joint powers also showed a delay in the peak ankle power and peak hip power in both the sagittal plane and frontal plane but again, outperforming the younger adults in time to peak power at the knee joint in the sagittal plane. This further supports our conclusion that the older adults may be using a different technique to execute the compensatory step.

Implications

These discoveries in our study can have further implications in fall prevention and fall rehabilitation programs in OA. Fall prevention programs and fall rehabilitation programs typically focus on maintaining or improving strength in patients but through the knowledge gained in our research, it shows that joint power (especially at the ankle) and timings of peak moments and powers could also play crucial roles in fall prevention. As a result, these programs should consider adjusting their exercise prescription and exercise parameters to promote an increase in joint energy absorption at the lower limb joints such as eccentric training. In regards to improving the time to peak moments and time to peak powers, we believe that the poorly timed peaks could be related to a misperception of the perturbation magnitude. By integrating the use of support braces and the emphasis of proprioception exercises into fall prevention or fall rehabilitation programs, these could possibly address timing deficits through the improvement of neuromuscular connections. We also believe that further investigation of the relationship between perturbation perception and the identified peak timing deficits should be considered to further our understanding of the age-related differences in the single-step compensatory stepping response.

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