New Insights into the Landing Phase of Reactive Stepping: Predictors of Control, Muscle Recruitment, Movement Restraints and Two-Step Responses

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Authors Declaration

I hereby declare that I am the sole author of this thesis. This is a true copy of the thesis, including any required final revisions, as accepted by my examiners.

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ABSTRACT

Preventing falls and fall-related injuries are public health challenges of the upmost importance for Canadians, particularly for older adults. Due to the severe consequences that can accompany a fall (e.g., traumatic brain injuries, hip fractures), studying age-related changes in balance control is an important avenue towards informing more effective fall-prevention interventions. As tripping is a common cause of falling in older adults, many researchers have studied reactive stepping after a simulated trip. While the majority of studies have focused on recovery at foot-contact (FC), researchers have begun to focus on the landing phase (or restabilisation phase), which occurs after the point of FC of the reactive step. This thesis adds to the burgeoning insights in this important area by addressing the following four general objectives: i) determine which individual characteristics are predictive of center of mass (COM) displacement during the reactive stepping and landing phases in young and older adults (Study 1); ii) determine how lower-limb muscle recruitment patterns during the landing phase compare to earlier phases of the reactive stepping response, as well as how lower-limb electromyography signals relate to each other during the landing phase (Study 2); iii) determine if wide stepping and restricted arm movement influence balance control during the landing phase (Study 3); and, iv) to quantify balance control after FC when participants responded with two reactive steps (Study 4). Reactive stepping was evoked via a tether-release paradigm.

In Study 1, it was found that for both young and older adults, regression models driven by specific tether-release metrics were stronger predictors of COM movement during the stepping and landing phases compared to general metrics, calculated separately from the tether-release trials (e.g., response time, range-of-motion, etc.). For Study 2, which quantified lower-limb electromyography (EMG), the peak timing and magnitude were generally slower (more variable) and smallest from the point of the maximum COM after FC to the end of the trial. The muscles which exhibited their highest peak magnitude during the landing phase were the biceps femoris of the step-leg, which was correlated with the peak medial gastrocnemius magnitude during landing, and the rectus femoris and tibialis anterior of the support-leg. Peak magnitudes suggest that the step-leg biceps femoris and medial gastrocnemius and support-leg rectus femoris (in continuation from the swing phase) and tibialis anterior are important

during landing, while the step-leg rectus femoris and tibialis anterior are important in the swing phase. Regarding the investigation of wide stepping and restricted arm movement in Study 3, wide stepping resulted in the largest medio-lateral (ML) and anterior-posterior (AP) body movement after FC, regardless of age group. Second, despite limited AP influence, restricted arm movement resulted in larger ML body movement after FC, compared to the preferred stepping condition. During Study 4, analyses of the first step revealed that during the two-step condition peak AP COM displacement after FC was increased, while peak ML COM displacement was decreased, for both loading conditions. With asymmetrical loading, first step lengths were larger during the one-step condition, while first step width was reduced over both stepping tasks with asymmetrical loading. During two-steps, peak AP extrapolated COM (xCOM) displacement after FC was larger in the second vs. the first step with asymmetrical loading, yet the first step resulted in greater ML xCOM displacement vs. the second step, regardless of loading. Interestingly, first step width was narrower than the second with asymmetrical loading. As hypothesized, peak xCOM displacement between the first and second steps was correlated.

As a whole, the results of these studies provide novel insights into the landing phase of reactive stepping. A consistent theme pertains to the potential ability of pro-actively training effective reactive stepping responses. The findings suggest that researchers and clinicians should consider task specificity if training reactive stepping responses. Furthermore, focus should be placed on the muscles which were their most active (and correlated) during landing, (i.e., the biceps femoris and medial gastrocnemius of the step-leg). The large stability margin observed during wide stepping suggests it can be a positive strategy for increased ML stability. Incorporating the arms into training would also be positive, as ML COM control decreased when the arms were restricted. Finally, multi-step balance control should not be inferred using single-step responses, as differences in COM\xCOM displacement existed between stepping tasks and step number. Further, xCOM correlations between the first and second step did not improve when the one-step responses were used for the first step metrics. To further enhance the evidence base in this area, future work could focus on characterizing landing phase control during more dynamic activities such as tripping during gait, ideally with participants who represent high-fall risk groups.

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LIST OF COMMON ABBREVIATIONS

AP = anterior-posterior

APA = anticipatory postural adjustment

APR = automatic postural response

BOS = base-of-support

BF = biceps femoris

BW = body weight

CNS = central nervous system

COM = center of mass

CR = cable-release

EMG = electromyography

FC = foot-contact

GRF = ground reaction force

MG = medial gastrocnemius

ML = medio-lateral

MoS = margin of safety

RF = rectus femoris

ROM = range-of-motion

TA = tibialis anterior

TUG = timed up and go

TO = toe-off

xCOM = extrapolated center of mass

TABLE OF DEPENDENT VARIABLES

Dependent Variable	Study	Method of Calculation
COM Displacement at FC	1	COM position at FC was referenced to the mean starting COM value (from the start of the trial (frame 1) to one frame before CR).
Peak COM Displacement after FC	1, 3, 4	Peak COM position after FC was referenced to the mean starting COM value (from the start of the trial (frame 1) to one frame before CR).
EMG Peak Timing (for each phase)	2	Peak timing was calculated as the time between the event of interest (i.e., Cable-release, TO, FC, etc.) and peak EMG magnitude (using the linear-enveloped signal).
EMG Peak Magnitude (for each phase)	2	Peak magnitude was calculated as the maximum value in a given phase (using the linear enveloped signal).
Stability Margin (at Peak COM Displacement after FC)	3	AP and ML stability margins were calculated by subtracting the COM position from the tip of the right big toe and the 5th metatarsal, in the AP and ML directions, respectively. Stability margins were calculated at the point of the maximum AP and ML COM positions after FC.
Step Length	3, 4	Step length was calculated using the COM of the foot. The difference in the AP position of the foot COM between FC and CR was calculated as the step length.
Step Width	3, 4	Step width was calculated using the COM of the foot. The difference in the ML position of the foot COM between FC and cable-release was calculated as the step width.
Peak xCOM Displacement after FC	4	Peak xCOM position after FC was referenced to the mean starting COM value (from the start of the trial (frame 1) to one frame before CR). xCOM was calculated as follows: $x + (\dot{x}/\sqrt{g/l})$

1. OVERVIEW, FRAMEWORK, OBJECTIVES AND GENERAL HYPOTHESES

1.1 Overview

Humans' abilities to stand, walk and run upright are ones that separate us from the majority of species on earth. However, they also makes us inherently unstable as approximately two thirds of our body mass is located in the upper third of our body (Winter et al., 1990), and bipedal ambulation reduces the size of our base-of-support (compared to quadrupeds). Unfortunately, increased age can result in a decreased ability to effectively maintain stability, increasing one's risk of falling (Maki & McIlroy, 2006; McIlroy & Maki, 1996; Nevitt et al., 1989; O'Loughlin et al., 1993; Tinetti et al., 1988).

Many researchers have reported differences in balance control/stability between young and older adults. For example, numerous studies have shown that older adults take multiple reactive steps after an external postural perturbation (Luchies et al., 1994; McIlroy & Maki, 1996; Schulz et al., 2005), which are slower and shorter in length (Thelen et al., 1997; Wojcik et al., 1999). Older adults also often step with a larger lateral component compared to young adults (McIlroy & Maki, 1996; Schulz, et al., 2005). Regarding gait, older adults exhibit a slower velocity (due to a reduced stride length), increased double-limb support time, a reduced push-off and a more flat-footed landing (Winter, et al., 1990). Balance impaired older adults have also been shown to exhibit excessive lateral momentum during gait, despite walking at a slower velocity (Kaya et al., 1998). Older adults also walk and terminate gait with an increased stride width (the former is moderately associated with falling (Maki, 1997)), and terminate their gait less frequently using one-step (Menant et al., 2009; Tirosh & Sparrow, 2004).

Towards further enhancing our knowledge of balance control, recently the landing phase (or the restabilisation phase) of stepping has received increased attention. The landing phase is the period when the stepping foot is back in contact with the ground, and has a direct influence on the kinematics of the center of mass (COM) after movement initiation (Singer et al., 2012), when individuals can apply (potentially) stabilizing forces against the support surface. In studies of older adults during voluntary and reactive stepping, the medio-lateral (ML) COM incongruity was larger in older adults, with greater variability (Singer et al., 2013, 2016). The COM incongruity refers to the difference between the

maximum ML COM position after foot-contact (FC) and the stable region of the ML COM (Singer, et al., 2013, 2016). Despite the novel findings, there is a need for research to understand the role of the landing phase, and more specifically, the role of body movement after FC. For example, it is possible that the alterations to the COM exhibited by the older adults could be pro-active strategies during the landing phase to aid in taking a subsequent step. Alternatively, the increased body movement after FC in older adults could be a form of dyscontrol due to age-related changes (Singer, et al., 2013).

Research on the landing phase of forward stepping suggests that in older adults, initial voluntary and reactive steps may not be as stabilizing as those used by young adults (Singer, 2012; Singer et al., 2014; Singer, et al., 2013, 2016). However, multiple aspects of the landing phase remain unexplored. First, it is unknown if balance control (i.e., COM displacement, velocity) during the stepping phase is related to balance control during the landing phase of reactive stepping. Further, it is unknown which factors are predictive of balance control during the landing phase, and if these factors differ between young and older adults. Second, and in-line with predicting balance control during the landing phase, it is unclear how lower-limb muscle recruitment patterns during the landing phase compare to other phases of the reactive stepping response. Third, while older adults often take wider steps and are more reliant on arm movement (compared to young adults), it is unclear if these strategies are beneficial for balance control during the landing phase. Lastly, while previous research has focused on the landing phase during single-step responses, the landing phase during two-step responses has received less attention. Understanding how balance control changes during the landing phase of two-step responses may provide important insight into the role of COM movement after FC, and will provide a springboard into the study of the landing phase during more dynamic activities, such as trip recovery during gait.

1.2   Framework: Neuromechanical Control and Sources of Potential Age-Related Differences

For humans to maintain an upright posture, their COM must remain positioned over the base-of-support (BOS) (Maki & McIlroy, 1997). However, external or internal perturbations may cause the COM to approach the limits of the BOS. Accordingly, to prevent a fall, two distinct categories of balance recovery strategies exist to bring the COM back over the BOS: fixed-support and change-in-support

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strategies (Maki & McIlroy, 1997). Fixed-support strategies may be used in response to small or moderate perturbations to one's balance. These strategies do not change the size of the BOS but instead aim to stop movement of the COM by generating muscle torques and causing rotation about the ankle, knee and hip joints (Maki & McIlroy, 2006). However, if an individual is exposed to a larger perturbation, it may be necessary to use a change-in-support strategy, which results in an increase in the size of the BOS (Maki & McIlroy, 2006). One specific example of a change-in-support strategy is reactive stepping, where the size of the BOS is increased based on where the stepping leg is placed. Recently, researchers have begun to focus on the landing (or restabilisation) phase of reactive stepping, which occurs after the point of foot-contact (FC) (King et al., 2012; Serrao et al., 2013; Singer, et al., 2014; Singer, et al., 2012, 2016).

The landing phase is important for balance recovery as it allows one to modulate their net center of pressure via the generation of step-leg ground reaction forces (GRFs) (Singer, et al., 2016). Age-related differences have previously been observed during the landing phase of reactive stepping, specifically in the peak COM displacement after FC (Singer, et al., 2016). Mechanically, these differences may be attributable to declines in the ability of older adults to generate appropriately directed GRFs (King, et al., 2012; Singer, et al., 2014), but not magnitude (King, et al., 2012). While generating GRFs of the appropriate magnitude may not be problematic for older adults, lower limb strength may be important during the landing phase. For instance, older adult multi-steppers tend to use a higher proportion of their available hip extension strength compared to single-steppers (Carty et al., 2012b). Others have also implicated the degeneration of leg extensor muscle-tendon units in age-related declines in forward stepping (Karamanidis & Arampatzis, 2007; Karamanidis et al., 2008). Interestingly, Pavol et al. (2001, 2002) also reported that low strength older adults may be at an increased risk of after-step falls, due to declines in knee and\or hip extension strength. Older adults also produce less step-limb joint power in response to a simulated-trip (Carty, et al., 2012b). Step length, width and movement time may also play an important role in landing phase control, and differences between young and older adults. Older adults are slower to react and move with slower and shorter steps (Carty et al., 2011; Lee et al., 2014; Luchies,

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et al., 1994; Schulz, et al., 2005; Thelen, et al., 1997; Wojcik, et al., 1999), which could each influence how far forward the COM is displaced after FC, during the landing phase.

In addition to the measures discussed above, neural factors such as muscle activation onset latency and automatic postural response (APR) timing\magnitude may also influence landing phase control. Automatic postural responses are muscle responses that typically precede the onset of a step (e.g., the generation of ankle moments via the triceps surae). If reactive stepping is the desired outcome, this can present a conflict in responses, as stepping would require a center of pressure displacement directed opposite of that generated by the early APR (Singer, 2012). Researchers have tried to better understand this potential conflict by studying reactive stepping after surface translations. Interestingly, APRs were always initiated, regardless of whether participants took a reactive step or not. Furthermore, the onset latency was the same. When steps were pre-planned, the magnitude of the tibialis anterior was reduced compared to unplanned stepping, or feet-in-place responses (McIlroy & Maki, 1993a). Researchers also observed the modifiability of the APR magnitude in a separate study where participants were asked to fall onto a mattress or to recover their balance after being released from a backward lean. Tibialis anterior onset latency was largely unchanged between the two conditions; however, the magnitude was significantly reduced in the fall trials (Weerdesteyn et al., 2008). This result suggests a similarity in the motor program between conditions, but that the motor program can be down-regulated as needed. Additional studies have reported similar findings on the consistency in timing of early APRs after release from forward leaning (Do et al., 1982), during reactive stepping, with and without constraints (McIlroy & Maki, 1993b), and also after being exposed to the same perturbation magnitude (McIlroy & Maki, 1995). Despite the consistent timing of early APRs during different tasks, researchers have reported that older adults exhibit delayed onset latencies in multiple muscles, during various balance recovery tasks. For example, older adults also exhibit delayed electromyographic (EMG) responses in all postural leg, hip, trunk and arm muscles after rotational perturbations (Allum et al., 2002), while anterior-posterior surface translations revealed delayed onset latencies in the medial gastrocnemius and biceps femoris in older adults (Tokuno et al., 2010).

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Lastly, occurring after the early APR, are anticipatory postural adjustments (APAs) which act to unload the stepping leg prior to toe-off. Research has shown that voluntary steps almost always occur with an APA preceding step initiation (Brunt et al., 1991). Unlike voluntary stepping, APAs are normally either absent or diminished in magnitude and effectiveness during reactive stepping (McIlroy & Maki, 1999; Rogers et al., 2001), which may have a negative effect on ML stability (McIlroy & Maki, 1999; Rogers, et al., 2001). One study did find that young adults were more likely than older adults to generate an APA prior to reactive stepping, however the magnitude of the APA was small, with no corresponding increase in lateral stability at the time of foot-contact (McIlroy & Maki, 1996). This was in agreement with separate (unpublished) research showing that APAs which do precede compensatory stepping are often too brief or too small to have a significant or functional influence on the lateral movement of the COM (McIlroy & Maki, 1996). Nonetheless, along with the other factors listed above, it remains possible that APAs could also contribute to the peak movement of the COM after foot-contact, thereby influencing landing phase control. A summary of all the neuromechanical factors discussed is depicted in Figure 1-1.

In light of the framework presented above, the current thesis focused on four specific areas which each encompassed some or all of the factors shown in Figure 1.1. Study 1 focused primarily on predictors of balance control during the landing phase, where factors such as response time, muscle strength and movement amplitude (e.g., step length and width) were examined, based on apriori decisions. Study 2 focused on the peak timing and peak magnitude of lower-limb muscle activity throughout the entirety of the reactive stepping response. Specifically, the peak muscle recruitment patterns during the landing phase may speak to factors occurring earlier in the reactive stepping response, such as muscle activation onset latency, or the size of the automatic postural response. If such factors which are present early in the reactive stepping response are ineffective in restraining the forward rotation of the body, conceivably greater muscle activation will be required during the landing phase. Study three focused on the effect of wide stepping and arm movement on landing phase control. Not only have these characteristics been reported in older adults for a variety of tasks, but it is plausible that the need for a wide step, or large arm movement after foot-contact could be related to factors during the early part of the response, or during the

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swing phase (e.g., muscle activation onset latency, automatic postural response timing and magnitude, anticipatory postural adjustment amplitude, movement time, etc.). Lastly, Study 4 focused on characterizing the landing phase during two-step responses, where any one of the above-mentioned factors could cause a participant to need a second reactive step. For example, if the automatic postural response is too small, the movement time too slow, and\or the accompanying first step too short, the COM may be pitched further forward resulting in the need for a second reactive step. Study-specific objectives and general hypotheses are presented below.

Neuromechanical Control – Sources of Potential Age-Related Differences During the Landing Phase

Ground Reaction Force Magnitude & Direction	Muscle Strength	Muscle Power
Step Length	Step Width	Movement Time (Step Time)
Muscle Activation Onset Latency	Automatic Postural Response	Anticipatory Postural Adjustment (Loading Phase)

Figure 1-1: A depiction of the potential sources of age-related differences during the landing phase of reactive stepping.

1.3 Objectives and Hypotheses

This thesis was comprised of four complementary studies outlined in Figure 1-2. Studies 1 – 3 were collected as part of one large experiment, while Study 4 was collected separately. In summary, the overarching objectives of this thesis were as follows:

1) Enhance our understanding of human balance control and recovery by assessing whether phases of forward reactive stepping are related in young and older adults. Focus was placed on COM displacement during the stepping phase (up to FC), and the landing phase (occurring after FC). Second, to determine which individual characteristics are predictive of COM displacement during both the stepping and landing phases, in young and older adults.

It was hypothesized that balance control at FC would correlate with balance control after FC, in both the young and older adults. Second, general measures would significantly predict COM displacement during the stepping and landing phases, in young and older adults. However, regressions incorporating specific measures (from the tether-release trials) would result in larger R and R^2 values compared to the general models, for each dependent variable and age-group.

2) Determine how lower-limb muscle recruitment patterns during the landing phase compare to earlier phases of the reactive stepping response. Specifically, to determine if peak muscle recruitment differs across phases of forward reactive stepping for muscles of the lower-limb. Second, to determine if the peak magnitude between muscles is related in a given phase of reactive stepping, in an attempt to gain insight into synergistic activity between muscles.

It was hypothesized that of the muscles examined, the peak timing and magnitude of the step-leg biceps femoris (due to its role as a hip-extensor) would be earliest and at its largest during the landing phase, from FC to the peak COM position after FC. Overall, for each muscle, the later aspect of the landing phase (after the peak COM) would have the latest peak timing, smallest peak magnitude, and the largest variability. During the period from FC to peak COM position after FC, the biceps femoris and medial gastrocnemius of the stepping leg would exhibit the largest positive correlation between peak magnitudes.

3) Determine whether wide stepping and restricted arm movement influence balance control during the landing phase of reactive stepping, in young and older adults.

It was hypothesized that wide stepping would result in the largest and most variable peak COM displacement after FC. Second, restricted arm movement would result in a larger and more variable peak COM displacement after FC, compared to the preferred stepping trials. Lastly, compared to the young adults, older adults would exhibit a larger and more variable peak COM displacement after FC during all experimental conditions. However, the between-group differences would be largest during the wide stepping and restricted arm movement conditions.

4) Quantify balance control after FC when participants responded with two reactive steps. A secondary objective was to quantify the effects of asymmetrical stance loading (pre-release) during two-step responses.

When comparing the first step, it was hypothesized that two-step responses would result in a larger peak COM displacement after FC, compared to one-step responses. Second, during the two-step responses, the second step (vs. the first step) would occur with a smaller peak COM displacement after FC. Asymmetrical loading would result in a smaller peak COM displacement after FC during the first step, but a larger peak COM displacement after FC during the second step.

Study 1: Predictors of balance control during the landing phase of reactive stepping in young and older adults

One larger data collection consisting of 40 young adult and 40 community-dwelling older adults

All participants completed:

-Maximal isometric hip extension task
-Maximal active hip flexion range-of-motion
-Maximal active hip abduction range-of-motion
-Voluntary stepping in response to an auditory tone

-2 quiet standing trials (60 seconds each)
-5 practice tether-release trials
-10 preferred stepping trials

Study 2: Characterizing peak lower-limb muscle electromyography during the landing phase of reactive stepping

-20 young adults (also participated in Study 1)

-Electromyography collected from the: rectus femoris, biceps femoris, tibialis anterior and medial gastrocnemius

Study 3: Influence of wide stepping and arm movement on balance control during reactive stepping: a focus on stability after foot-contact

-20 young adults and 16 older adults (also participated in Study 1)

-5 wide stepping trials
-5 restricted arm movement trials

Study 4: Quantifying balance control after foot-contact during two-step responses

One separate data collection which involved 18 young adults

All participants completed:

-2 quiet standing trials (60 seconds each)
-5 practice tether-release trials
-Single-step trials, right (dominant) leg first
-Two-step trials, right (dominant) leg first

For both of the reactive stepping blocks, participants completed one block where they started with the body weight evenly distributed between legs, and one block with an uneven body weight distribution between legs. The uneven distribution consisted of a 60%/40% split, where the left (non-stepping) leg was initially loaded to a larger degree.

Figure 1-2: A graphical overview of the four studies which comprised this thesis. The two arrows leading from Study 1 to Studies 2 and 3, respectively, represent the fact that Studies 2 and 3 were collected at the same time as Study 1 (i.e., part of one large data collection). Study 4 was collected separately, and involved only young adult participants.

2. REVIEW OF LITERATURE

2.1   Overview – Falls in Older Adults

Globally, the number of people who are of the age 65 years and older is growing at a faster rate than any other age group (WHO, 2007). In Canada specifically, the older adult cohort accounted for 15% of the population in 2011, and is expected to increases to 25% by the year 2036 (Statistics Canada, 2010). Due to the fact that the proportion of older adults living in Canada is increasing, we must devote our attention to the issues that these people face in living their daily lives. Accordingly, falls are a major public health concern and the leading cause of injury in older adults above the age of 65, as approximately one-third of community-dwelling older adults experience a fall annually (Raina et al., 1997; Tromp et al., 2001). Researchers have suggested that 20% of falls in older adults will result in serious injury such as hip and intertrochanteric fractures and traumatic brain injuries (Alexander et al., 1992; Gilasi et al., 2015). Furthermore, 77% of all injury hospitalizations in older adults are due to fall related injuries (Johnson et al., 2015; PHAC, 2005). Economically, this places a huge burden on our health-care system as an estimated 2 billion dollars is spent annually on fall related injuries in older adults (SMARTRISK, 2009). Even in the absence of a fall-related injury, falling once increases the likelihood of falling again and the fear of falling itself can cause mobility and independence limitations (O'Loughlin, et al., 1993; Tinetti et al., 1994). Therefore, it is of paramount importance that we conduct research aimed at characterizing falling and balance control which can inform fall reduction strategies.

2.2   Overview – Reactive Stepping in Response to an External Perturbation

In order to prevent a fall, two different types of balance recovery strategies may be employed to bring the center of mass (COM) back over the base-of-support (BOS), and maintain upright posture (Maki & McIlroy, 1997). Fixed-support strategies commonly occur in response to a small or moderate perturbation to one's balance. These strategies do not change the size of the BOS, but instead the goal is to control movement of the COM by generating oppositely-directed (to the rotation of the COM) muscle torques, causing rotation about the ankle, knee and hip joints (Maki & McIlroy, 2006). When a larger

perturbation is experienced, or when one's BOS is perturbed (e.g., their feet) via a trip or slip, it may be necessary to rely upon a change-in-support strategy. A change-in-support strategy results in a large increase in the size of the BOS either through stepping or the grasping of an object (e.g., handrail) for support (Maki & McIlroy, 2006). Consequently, such a strategy allows for a much larger range of COM motion to occur before it is brought to rest (Maki & McIlroy, 2006).

Although the size of a perturbation influences one's balance recovery strategy, an individual's balance capabilities will also determine whether they use a fixed or change-in-support strategy. For example, older adults tend to rely more heavily on change-in-support strategies in order to recover their balance. Video surveillance of a geriatric facility showed that reactive stepping was quite common in response to a loss of balance (Holliday et al., 1990), as in 45% of cases individuals tried to take a step prior to the fall. Similarly, Yang et al. (2013) reported that observable attempts to recover balance via reactive stepping were noted in 42% of falls. Interestingly, even when participants are unsuccessful at recovering their balance via reactive stepping, the act of taking a step may still help to reduce the risk for hip fracture by increasing the time to pelvis impact, increasing the interval between hand and pelvis impact and decreasing the impact velocity of the pelvis (Feldman & Robinovitch, 2007).

Regarding response prevalence, multiple laboratory studies have found reactive stepping to be common in response to postural perturbations. For instance, using perturbations which occurred in multiple directions, Hsiao and Robinovitch (1998) observed reactive stepping to be the predominant balance recovery response for young adults. Additionally, when participants did experience a fall, a failed reactive step was still observed in all but one fall trial (Hsiao & Robinovitch, 1998). Further, reactive stepping was prevalent even when participants were exposed to small-magnitude perturbations (Hsiao & Robinovitch, 1998; Luchies, et al., 1994; Pai et al., 1998; Rogers, et al., 2001). However, reactive steps can also be adapted to the initial disequilibrium torque, allowing for faster and larger reactive steps, for larger perturbations (Do, et al., 1982). As reactive stepping is a common balance recovery response in young adults, it is not surprising that this response is also frequently used\attempted by older adults, as shown by the studies of real-life falls in the paragraph above.

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To better understand why reactive steps are initiated, many researchers have quantified dynamic balance by considering position-velocity relationships between the COM and BOS, moving beyond just COM position measures (Carty, et al., 2011; Hof et al., 2005; Pai et al., 2000; Pai & Patton, 1997; Pai, et al., 1998). Specifically, one study compared static (i.e., dependent on COM displacement) and dynamic (i.e., where the initial position and velocity of the COM and BOS were used as inputs) inverted pendulum models. Overwhelmingly, the dynamic model better predicted the need to take a reactive step in response to a support surface translation (i.e., 71% vs. 11% of stepping responses were predicted correctly for the dynamic and static models, respectively) (Pai, et al., 2000). This result was similar to a study when the COM was perturbed via waist-pulls. In that study, the dynamic model predicted stepping with an accuracy of 65%, compared to 5% for the static model which did not consider COM velocity (Pai, et al., 1998). This demonstrated that the central nervous system (CNS) must react to and control the COM velocity as well as the COM displacement in order to effectively maintain stability without stepping (Pai, et al., 2000). Detecting and controlling both the COM displacement and velocity is difficult; therefore it is not surprising that reactive stepping is a very common response, even when participants are told not to step (McIlroy & Maki, 1993b). Accordingly, more recent models developed to predict if a balance perturbation can be recovered from using a single recovery step (as well as estimating the characteristics of the most efficient recovery step – shortest and fastest step) have incorporated and considered velocity effects about the COM, in the form of the extrapolated COM (Vallee et al., 2015). However, the decision to initiate a reactive step may also reflect non-biomechanical factors such as a fear of falling, misperception of the current state of instability or an anticipation of the forthcoming instability (Pai, et al., 2000). Due to the importance and frequent occurrence of reactive stepping, many researchers have investigated how reactive stepping changes in older adults.

2.3   *The Effect of Age on Reactive Stepping*

The biomechanics of reactive stepping have been widely studied in older adults. Wojcik et al. (1999) observed that 50% of older women were unable to recover their balance with a single-step after being released from the smallest of the imposed forward-lean perturbations. This inability may have been

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due to limitations in the maximum speeds at which the older women moved their swing foot (Wojcik, et al., 1999). Similarly, Thelen et al. (1997) observed that the maximum lean angle from which older adults could recover from using a single reactive step was significantly smaller compared to young adults. Similarly, multi-step reactions in older adults have also been reported by Schulz et al. (2005) and Luchies et al. (1994) when responding to larger perturbations, Mille et al. (2013) when responding to waist pull perturbations, Dijkstra et al (2015) after anterior-posterior (AP) surface translations, and McIlroy and Maki (1996), where multiple stepping was almost twice as common in older adults, compared to young adults (63% vs. 35% of trials) after an AP surface translation. Further, the number (absolute and relative) of multiple steppers who experienced a fall in a 12-month follow-up period was greater compared with single-steppers (Carty et al., 2014). Not only do older adults often respond with multiple reactive steps, but anterior stepping thresholds are also reduced with age (Crenshaw & Grabiner, 2014).

Older adults who do take multiple-steps also step with a significantly shorter and faster initial recovery step, and adopt more trunk flexion throughout recovery (Graham et al., 2014). Shorter step lengths among multiple steppers were also reported by Cronin et al. (2013). Accordingly, older adult multiple steppers exhibit a smaller margin of safety (MoS) at foot-contact and at the maximum knee flexion angle after foot-contact, compared older adult single-steppers. Both the trunk flexion angle at foot-contact and step length correlated with the MoS at foot-contact (Carty, et al., 2011). Reduced lower-limb isometric strength (i.e., especially of the hip flexors and knee extensors, which are needed for taking a long step) is also associated with an increased odds of requiring multiple steps, vs. a single-step in older adults (Carty et al., 2012a). In contrast, others have observed no differences in leg extensor strength (isometric ankle plantarflexion and knee extension) or tendon stiffness between older adult single and multiple steppers (Arampatzis et al., 2008). Lastly, some researchers have suggested an important role for the hip joint, as older adult multi-steppers use a higher proportion of their hip extension strength but produce less knee and ankle joint peak power during stepping (Carty, et al., 2012b).

Deficits have also been noted in the speed and length of reactive steps, where older adults take slower (velocity) steps after a forward or backward platform perturbation (Lee, et al., 2014). Both Thelen

et al. (1997) and Wojcik et al. (1999) noted that older adults had slower reaction times compared to young adults, while Thelen et al. (1997) observed that older adults tended to take shorter reactive steps at any given forward-lean angle. The latter finding agreed with the work of Schulz et al. (2005) for backwards stepping and Luchies et al. (1994), who observed older adults to take shorter reactive steps, with a reduced height from the ground. Luchies et al. (1994) also observed that older adults landed their first reactive step earlier, with a shorter step duration. At foot-contact and up to the point of maximum knee flexion after foot-contact (i.e., termination of downwards body motion), older adults exhibited a shorter anterior BOS and smaller stability margin after a forward tether-release (Karamanidis, et al., 2008), possibly due to reduced lower-limb strength (Karamanidis, et al., 2008). Madigan and Lloyd (2005b) also reported that older adults exhibit reduced hip flexion, knee flexion and extension as well as ankle plantar flexion velocity when taking forward reactive steps. Similarly, Carty et al. (2011) found older adults to take shorter reactive steps, exhibit increased trunk flexion, smaller peak knee flexion angles and a smaller MoS at foot-contact vs. young adults.

In young and older adults, differences also exist for measures of medio-lateral (ML) stability. For instance, single-step trials were least frequently taken by older adults for purely lateral perturbation directions (Mille, et al., 2013). When forced to rely on a complicated cross-over lateral reactive step, there was also a 16-fold increase in inter-limb collisions for older adults, compared to young adults (Mille, et al., 2013). Along with using a laterally-directed balance task to examine ML balance control, lateral instability has also been observed in response to AP perturbations. For instance, in response to backward platform translations (causing forward stepping) older adults commonly exhibited "same-leg" multiple-step responses, where the "same-leg" reactions were almost exclusive to the older adults. In older adults, these responses occurred in over 30% of trials, often times featuring a second step where the foot moved laterally as much as 28 cm (McIlroy & Maki, 1996). Conversely, laterally directed second steps were only observed in four trials (8%) for the young adults. Lateral stepping may reflect an impaired ability to control a ML instability that arises after the initial foot-contact, subsequent to the onset of additional steps (McIlroy & Maki, 1996).

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Focusing on the initial step, recent research has shown that in older adults, the initial reactive step resulted in an increased ML ground reaction force (GRF) component, compared to young adults (King, et al., 2012). This suggests that ML instability in older adults may not be solely an issue of magnitude, but may arise from an inability to adequately control and direct the line of action of the net GRF, as the increased ML GRF component must be offset with an appropriately sized vertical force. Similarly, Rogers et al. (2001) also examined forward-directed reactive steps and observed that older adult fallers had a significantly greater and faster ML body motion toward the stepping side at first step contact and a more laterally directed step placement compared to young adults and older adult non-fallers. The authors suggested that the older adult fallers may have used a wider step in a compensatory manner, to account for the ML instability that likely developed from the point of foot lift-off to initial step-contact (Rogers, et al., 2001). However, no data was presented to quantify stability during the landing phase.

Impaired ML control may also exist in older adults during backward-directed reactive stepping, as compared to young adults, older adults show a larger lateral component to their reactive steps (Schulz, et al., 2005). Older adults also use more laterally-directed steps after a backwards directed slip (Troy et al., 2008). This may be a compensatory mechanism to account for increased lateral instability. The COM of the older adults moved more laterally before step lift-off and their feet also travelled more laterally during the first 0.1 seconds of their backwards compensatory steps. By directing their first step more laterally, the older adults reduced the usefulness of that step in resisting the posterior perturbation, but increased their lateral stability with the resulting BOS. Interestingly, when multiple step trials were compared to single-step trials, the initial posterior steps of the multiple step trials exhibited an increased lateral placement and increased initial lateral foot velocities. This suggests that multiple steps may not be as efficient as one long AP step in dissipating the backward-directed perturbation force, but appear to provide a more stable, conservative recovery strategy, providing stability in both the AP and ML directions (Schulz, et al., 2005).

Conservative strategies are also a recurring theme when one reviews the literature on ML stability during gait and gait termination in older adults. For example, when instructed to walk along a narrow

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path, marked by lines on the floor, older adults walked in a conservative manner by taking wider steps compared to the young adults (Schrager et al., 2008). Regarding gait termination, Tirosh and Sparrow (2004) postulated that upon the completion of the first step, if AP stability is secured but ML stability remains to be consolidated, the prevalence of a second step in older adults may occur to ensure that ML stability is achieved. This could explain why older adults often performed two-step stopping even when, after the first step, they were already within the stability region predicted by Pai and Patton (1997), as these authors modeled only AP stability. An increased stride width in older adults has also been reported by Menant et al. (2009) for rapid gait termination after an auditory cue.

While older adults do show reactive stepping deficits, when compared to young adults, they can adapt to repeated forward tether-release perturbations and improve their subsequent AP stability. Reactive stepping improvements have been inferred through a reduced number of reactive steps, a larger BOS at foot-contact, a reduced COM position at foot-contact, and an increased rate of BOS displacement from toe-off to foot-contact (Carty et al., 2012c). Researchers have also noted a reduced anterior COM position and velocity (i.e., resulting in a greater MoS) at both foot-contact and the maximum knee joint flexion angle after foot-contact (Barrett et al., 2012). No improvements were observed in the ML MoS with repeated trials/exposure (Barrett, et al., 2012). The fact that older adults can improve their reactive stepping has positive implications for training interventions to reduce the risk of falling in older adults. Of concern is that recent research has reported that single-step balance recovery from a forward-lean perturbation (simulated-trip) can begin to decrease as early as 51 years of age, where the maximum lean angle decreased below one standard deviation of the mean value for young adults 18 years of age (Carbonneau & Smeesters, 2014). As such, interventions to offset age-related declines in reactive stepping and other balance recovery responses are of the upmost importance, and may have positive benefits for middle-aged adults as well as older adults.

2.4   *Implications Resulting from a Loss of Balance*

Due to the severe consequences that can accompany a fall, studying balance control in older adults is very important. One of the injuries that can result from a fall is a traumatic brain injury (TBI). In

a study of TBIs, falls accounted for 71% (15 of 21) of injuries in older adults, 65 years of age and older (Pickett et al., 2001). Over half (50.3%) of all fall-related deaths in older adults 65 years of age and older can be attributed to fall-related TBI (Thomas et al., 2008). Of further concern is the fact that the risk for fall-related TBI continues to increase as one ages. Specifically, persons over the age of 85 are hospitalized for fall-related TBI over twice as often as those aged 75–84, and over 6 times as often as those aged 65–74 (Coronado et al., 2005). Linking fall mechanics to head impacts, videos of real-life falls in long-term care showed incorrect weight shifting (34%) and tripping (25%) to be the two highest causes of falls in which head impact occurred. The probability of head impact was significantly associated with initial fall direction and landing configuration (Schonnop et al., 2013). For falls that were initially directed forward, the odds ratio for head impact was at least 2.7-fold greater than for falls directed backwards, sideways or straight down. Further, the odds ratio for head impact was 2.5-fold larger for falls involving a forward or sideways landing configuration than a backward landing configuration (Schonnop, et al., 2013). While it is intuitive that forward-directed falls increase one's risk of sustaining a head impact, forward (along with sideways) directed falls also increase one's risk of sustaining a hip impact.

Another serious injury that can result from a fall is a hip fracture. In older adults, suffering a hip fracture was associated with a greater than 2-fold increase in the likelihood of death, a 4-fold increase in the likelihood of requiring long-term nursing facility care, and a 2-fold increase in one's probability of entering into a low-income socio-economic status (Tajeu et al., 2014). Regarding fall mechanics, hip fracture risk in older adult females increased 3.3-fold for those who fell sideways and approximately 30-fold for those who landed on or near the hip (Nevitt & Cummings, 1993). In older adult males, a sideways fall increased the risk of hip fracture by 3.2-fold, while hitting the hip and or thigh when falling increased the risk for hip fracture by almost 50-fold (Schwartz et al., 1998). Similarly, a prospective study of hip fracture cases in a LTC facility revealed that of those participants who suffered a hip fracture due to a fall, they were more likely to have fallen sideways (odds ratio = 5.7). The authors concluded that sustaining a fall to the side was an important and independent risk factor for suffering a hip fracture (Greenspan et al., 1998). However, not all impacts occurring to the side of the body are due to sideways-directed falls.

Similar to the findings of Schonopp et al (2013) regarding head impact, Yang et al. (2015) found that the odds for hip impact were 4.2 to 7.9-fold greater for falls initially directed forwards or sideways compared to falls directed backwards or straight down.

Due to the serious consequences which can result from a fall, such as a TBI or a hip fracture, it is important to be able to safely simulate balance perturbations in a lab setting. One experimental technique which can be used to accomplish this goal is referred to as the tether-release, cable-release, or lean-and-release (Hsiao-Wecksler, 2008), and will be described in section 3.3.1 of the document. Simply, the tether-release method simulates body configuration at the onset of a trip, which is a common cause of falls among older adults in long-term care (Robinovitch et al., 2013). Although the tether-release primarily perturbs the participant forward, a sideways perturbation will also be introduced if a reactive step is required (e.g., particularly during single-leg support). Due to the importance of reactive stepping for preventing a fall, we must understand all phases of the stepping reaction, including the landing phase (or restabilisation phase) which occurs after foot-contact. Historically, this phase has received less attention compared to the stepping phase (from toe-off to foot-contact), but recent investigations into the landing phase have provided novel insights into reactive stepping control and how it changes with age.

2.5   *The Landing Phase of Stepping*

2.5.1   *Definition and Current Knowledge*

Recently, researchers have begun to investigate the landing phase (or the restabilisation phase) of forward stepping, which occurs after heel-contact. Studying balance after foot-contact is very important, as research has shown incorrect weight shifting to be the largest cause of falls among older adults in long-term care (Robinovitch, et al., 2013), which shows that stability is not guaranteed just because both feet are in contact with the floor. Mechanically, the requirements of the stepping leg are not expected to become critical until after landing, when forces and moments must be generated to counteract the angular momentum of the body (Pijnappels et al., 2004). Accordingly, if placed correctly, in front of the body's COM, the recovery limb can generate a moment that counteracts the body's forward rotation (Grabiner et al., 1993). According to Singer et al. (2012) this phase is particularly important for the maintenance of

dynamic stability because it may have the most direct influence on the kinematics of the COM after movement initiation when the swing phase is complete and the foot is back in-contact with the ground. Separate studies have also supported this notion. For example, Schulz et al. (2005) used waist pulls to study reactive stepping in the anterior and posterior directions in young and older adults, as well as balance-impaired older adults. Compared to the young and older adults, balance-impaired older adults were less effective at attenuating their linear momentum during the landing of the first step (inferred through AP COM-BOS distance), which may have contributed to their need for multiple steps. Schulz et al. (2005) suggested this indicated that critical balance impairment-related decrements in compensatory stepping occur after the landing of the first step. Challenges with balance control during the landing phase may also be evident from the finding of multi-step responses when older adults try to regain their balance using one-step (Luchies, et al., 1994; McIlroy & Maki, 1996). The need for additional steps may arise from a difficulty in the regulation of the position and velocity of the COM within the BOS, from the time beginning specifically after foot-contact occurs (Singer, et al., 2012).

Accordingly, researchers have begun to focus specifically on balance control during the landing phase of a forward step (after foot-contact). Results revealed that older adults use larger laterally directed landing phase ground and ankle reaction forces (King, et al., 2012), and show increased variability in frontal-plane balance control during rapid stepping (Kurz et al., 2013). Madigan and Lloyd (2005a) examined landing phase peak joint torques during single-step recovery, and observed a consistent pattern of joint torques between young and older participants, but older adults tended to use larger peak extensor torques at the hip and ankle (Madigan & Lloyd, 2005a). A separate group examined age-related changes in recovery after a forward fall, along with the influence of running experience. Young adults generally had increased muscle strength, increased tendon stiffness and better dynamic balance (i.e., increased BOS and MoS) compared to older adults, and running experience was linked to the ability to recover using a single-step (vs. non-active participants) (Karamanidis & Arampatzis, 2007; Karamanidis, et al., 2008). While these studies did capture the landing phase (or the stance phase as the authors referred to it), the analysis was limited to 400 ms after foot-contact (Karamanidis & Arampatzis, 2007; Karamanidis, et al.,

2008). The 400 ms cut-off was proposed to align with the termination of downward body movement after stepping (encompassing the first minimum angle at the knee after foot-contact); yet, data from the current projects show that maximum downward body position (COM) is not always complete within 400 ms after foot-contact, nor has the maximum AP or ML COM position after foot-contact always been reached. Separate studies have examined balance recovery up to the point of the maximum knee flexion angle after foot-contact (Arampatzis, et al., 2008; Arampatzis et al., 2011; Barrett, et al., 2012; Carty, et al., 2011); however, this method does not encompass AP or ML COM movement which may occur after the termination of downward body movement. To analyze the landing phase thoroughly, researchers should use a longer period after foot-contact.

In line with the previous paragraph, researchers have collected data for upwards of 10 seconds after the start of data collection, to allow for the landing phase to be adequately characterized (Singer, 2012). During a study of dynamic stability control during volitional stepping, researchers observed that overshoots in the final COM position were quite prevalent, occurring in 77% (AP) and 68% (ML) of all trials when participants stepped with their preferred step length and width. Overall, less than 30% of trials contained no incongruity (i.e., difference) between the peak and final COM position (Singer, et al., 2012). Although both AP and ML overshoots were observed, the authors focused on the ML COM overshoots. As the ML COM overshoots (toward the lateral BOS limit) occurred during voluntary stepping in a group of young participants with no known health issues, the authors initially suggested that the ML COM incongruity may serve a functional role (Singer, et al., 2012). During voluntary stepping, this was supported by the finding that the magnitude of the ML COM incongruity in young adults increased with a more conservative, wider step (Singer, et al., 2012). If the incongruity was a form of dyscontrol or an "error", one would expect that taking a wider step would reduce the size of this COM "error" in the ML direction. Instead, the incongruity may simplify reactive control after foot-contact, as greater than expected AP or ML movement of the COM could place an increased emphasis on the stepping limb for stability restoration. In theory, difficulties in COM control, resulting in a larger than expected forward or lateral COM movement after foot-contact would only require an increase in the force applied by the

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stepping limb and\or the initiation of an additional forward or lateral step to regain stability (Singer, et al., 2012). These characteristics are typical of the stepping responses observed in older adults in response to anterior postural perturbations (McIlroy & Maki, 1996).

However, this notion was challenged when researchers observed older adults to respond with a larger and more variable ML COM incongruity, regardless of whether older adults took voluntary steps of preferred length\width, reduced width, preferred speed or rapid speed (Singer, et al., 2013). Furthermore, when using forward reactive steps to respond to tether-release perturbations, older adults also exhibited an increased ML COM incongruity, regardless of whether they stepped with a preferred step placement (length\width) or with a reduced step width (Singer, 2012; Singer, et al., 2016). If alterations to the COM exhibited by the older adults are to be considered pro-active strategies during the landing phase, then purposely allowing the COM to travel closer to the lateral limits of the BOS would be counterproductive to the objective of maintaining stability. It seems more likely that the increased overshoot (and increased variability) in older adults (Singer, 2012; Singer, et al., 2013, 2016) was a function of dynamic stability dyscontrol (Singer, et al., 2013). Interestingly, it has been observed that in older adults, decreased strength may be linked with an increase likelihood of sustaining an after-step fall (i.e., a fall occurring at least 470 ms after recovery step contact) (Pavol, et al., 2002).

Overall, research on the landing phase of forward stepping suggests that in older adults, initial voluntary and forward reactive steps may not be as stabilizing as in young adults (Singer, 2012; Singer, et al., 2013, 2016), perhaps contributing to the use of multiple steps by older adults during many different activities. However, there are currently multiple research gaps which must be addressed to allow for a more comprehensive understanding of the landing phase, and the differences which exist between young and older adults.

2.5.2   *Research Gaps*

Although existing research on the landing phase has provided important and novel insight into this understudied phase of balance control, certain gaps still exist in the research. In older adults, impairments have been observed in the speed (slower) and length (shorter) of reactive steps (Lee, et al.,

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2014; Luchies, et al., 1994; Schulz, et al., 2005; Thelen, et al., 1997; Wojcik, et al., 1999). Regarding the role of leg-strength during reactive stepping in older adults, some studies suggest leg strength is important (Carty, et al., 2012a; Carty, et al., 2012b; Karamanidis & Arampatzis, 2007; Karamanidis, et al., 2008), while others suggest leg strength may not be very important (Arampatzis, et al., 2008; Grabiner et al., 2005). Graham et al. (2014) explored muscle force contributions during forward reactive stepping, in young and older adults, but did not analyze beyond foot-contact. Graham et al. (2015) explored biomechanical predictors of the maximum forward lean magnitude which participants could recover from with a single-step. While useful, this metric does not provide specific insight into balance control (i.e., COM displacement, stability margins, etc.). Therefore, to further develop the literature, research is needed to: 1) determine if balance control during the stepping phase and landing phase is related; and, 2) determine whether landing phase (and stepping phase) performance can be predicted by general and specific measures of reaction time or movement time, range-of-motion\movement size and leg strength. It is important to understand if balance control during the stepping and landing phases is correlated, as it would suggest a potential for transfer effects when training a specific phase of reactive stepping. It is known that the support-leg can provide enough time and clearance for proper positioning of the step-leg, and reduce the angular momentum during push-off, reducing the step-leg demands (Pijnappels, et al., 2004). Despite these results, it is unclear if balance control (i.e., COM displacement, velocity) between phases is correlated. Previous research has advocated for both the COM displacement and COM velocity when studying balance control, but also for predicting the need to initiate a reactive step (Hof, et al., 2005; Pai, et al., 2000). Second, in predicting balance control during the landing phase of reactive stepping, it would also be ideal for predictor variables to be easily measured and assessed by clinicians. For example, if lower-limb strength was shown to be a strong predictor of landing phase performance, such information could be incorporated into interventions, in the form of resistance training, to improve balance control in older adults. These research gaps will be addressed in Study 1 (Chapter 3) (refer to thesis framework in Figure 1-2).

Second, multiple researchers have reported that older adults exhibit an impaired timing of muscle activity when responding to postural perturbations. In a study of older adult single vs. multiple steppers, single-steppers recruited a larger proportion of the available motor unit pool during balance recovery (higher peak normalized electromyography (EMG) in 6 of 7 step-leg muscles), and also took longer steps than multiple steppers (Cronin, et al., 2013). Specific balance metrics (e.g., COM displacement, stability margins) were not quantified, and the study focused only on up to foot-contact, meaning the landing phase was not analyzed. Similarly, in response to feet-in-place rotational perturbations, older adults exhibited delayed EMG responses in all postural leg, hip, trunk and arm muscles, potentially compensated for by enhanced later responses (Allum, et al., 2002). In response to anterior-posterior surface translations, older adults also responded with delayed EMG onset latencies in the medial gastrocnemius and biceps femoris (Tokuno, et al., 2010). Older adults also recruit fewer swing leg muscles, resulting in less extensor torque (Tirosh & Sparrow, 2005). These examples show that impaired muscle recruitment has been observed in older adults across multiple types of balance recovery tasks\perturbations. However, no study to date has assessed lower-limb muscle recruitment patterns (peak timing and magnitude) during the landing phase, and compared these patterns to earlier phases of the reactive stepping response. Further, no study has assessed between-muscle relationships in peak timing and magnitude within each phase. Therefore, prior to assessing landing phase recruitment patterns in older adults, a baseline of control must first be established in young adults. Moving forward, understanding how lower-limb muscle recruitment differs between reactive stepping phases will allow researchers and clinicians to target phases of reactive stepping, by training a muscle(s) which is at its most active during the specific phase. Further, understanding if separate muscle recruitment patterns correlate within a phase will also allow for specific training, by targeting (e.g., strength training) the muscles with correlated recruitment patterns in a given reactive stepping phase. An investigation of muscle recruitment during the landing phase will be addressed in Study 2 (Chapter 4) (Figure 1-2).

Third, it is unclear how specific movement characteristics influence balance control during the landing phase. For example, older adult fallers use more laterally directed forward-reactive steps

compared to young adults and older adult non-fallers (Rogers, et al., 2001). In response to AP platform translations, older adults were more likely than young adults to take laterally-directed additional steps (McIlroy & Maki, 1996). Likewise, larger lateral steps have been observed during backward-directed stepping (compared to young adults) (Schulz, et al., 2005; Troy, et al., 2008), gait (Dean et al., 2007), narrow gait (Schrager, et al., 2008) and gait termination (Menant, et al., 2009). Stride width variability has also been found to be larger in older, compared to younger, adults (Dean, et al., 2007; Grabiner et al., 2001). An increased stride width in older adults has also been shown to have a moderate association with both falling and fear-of-falling (Maki, 1997). Approximately 70-85% of the gait cycle occurs in the single-leg phases (Maki, 1997), therefore older adults may be forced to adopt a wider stride to capture the COM as it falls sideways during single-limb support before effectively moving into double-limb support. Older adults are also more reliant on the use of their arms for assistance during balance recovery after a postural perturbation (Maki et al., 2000), despite being unable to initiate arm movements as rapidly (Allum, et al., 2002; Maki et al., 2001; Mansfield & Maki, 2009; Weaver et al., 2012), or to the same extent as the young (Allum, et al., 2002). However, the influence of arm movement on landing phase control has not yet been studied. Understanding how wide stepping and restricted arm movement influence balance control during the landing phase has important implications for training reactive balance control. If wide steps or arm movements have a positive effect on landing phase control, these actions can be emphasized in clinical settings. From a basic science perspective, studying landing phase control during wide stepping (specifically in the ML direction) will provide important information on the role of body movement after foot-contact (e.g., is the COM overshoot proactive or poor control). The influence of wide stepping and restricted arm movement on balance control during the landing phase will be addressed in Study 3 (Chapter 5) (Figure 1-2).

Lastly, throughout day-to-day life, the most common types of perturbations initiated are voluntary (e.g., walking, turning, reaching or bending). One such internal perturbation occurs during gait termination, which requires the CNS to predict the future and final position of the body's COM (Winter, 1995). This may be particularly difficult for older adults, as they often terminate gait using more than

one-step (Menant, et al., 2009; Tirosh & Sparrow, 2004, 2005). However, it is unclear if this need for multiple steps is real or "perceived" as 86% of the older adults' two-step responses occurred within the predicted stability region prior to them taking the second step (Tirosh & Sparrow, 2004). Recently, the landing phase of single-step responses has become a focus of researchers (Singer, et al., 2014; Singer, et al., 2012, 2013, 2016). Accordingly, it is of interest to study the landing phase of reactive stepping when participants use two-steps to recover their balance because it has implications for tasks occurring after a large postural perturbation, such as tripping while walking. As an initial step, researchers must characterize the landing phase during two-step responses in young adults to establish a baseline of control and to determine if single-step responses are similar to those observed during multi-step scenarios, in terms of body movement after foot-contact and foot-placement (step length and width). Effects of asymmetrical loading between legs should also be assessed as during dynamic activities involving gait, one's body weight is unevenly distributed throughout the various phases. Further, these results could carry implications for those individuals who exhibit loading asymmetries during standing\leaning (e.g., stroke patients). Studying the landing phase during two-step responses will provide important information on the role of body movement after foot-contact of a reactive step. As such, landing phase control during two-step responses, as well as the influence of asymmetrical leaning will all be addressed in Study 4 (Chapter 6) (Figure 1-2). Future work should progress to studying the landing phase in more dynamic activities, moving beyond responses evoked from a stationary starting position.

Overall, studying the landing phase of reactive stepping is very important. Research is needed to further understand this phase, so researchers can progress to studying the same phase in individuals at an elevated risk of falling compared to community-dwelling older adults. For example, frail older adults, individuals with Parkinson's disease, or stroke patients. In response to a trip, the act of taking of step alone does not guarantee that one will not fall. One must continue to control their body movement after the point of foot-contact. Recall, in long term care, frail-older adults were very susceptible to falls due to tripping while walking, but also due to incorrect weight shifting (Robinovitch, et al., 2013), which shows the importance of body control even when both feet are in contact with the support surface.

3. STUDY ONE – PREDICTORS OF BALANCE CONTROL DURING THE LANDING PHASE OF REACTIVE STEPPING IN YOUNG AND OLDER ADULTS

3.1 Chapter Overview

Research on the landing phase of reactive stepping, occurring after foot-contact (FC), has revealed control deficits in older adults. Currently, it is unknown if balance control during the stepping and landing phases is related, and if individual characteristics are predictive of balance control during reactive stepping. The purposes of this study were to determine if elements of balance control during the stepping and landing phases are related, and to assess the predictive value of models consisting of general and specific (from the tether-release trials) predictors. Forty young adults and 40 older adults participated in the study. A tether-release paradigm was used to evoke 10 reactive stepping responses. Dependent variables were center of mass (COM) displacement at FC and peak COM displacement after FC. Predictor variables were: maximal isometric hip extension strength; maximal active hip flexion and hip abduction (general) and normalized step length and width (specific); response time, when stepping to an auditory tone (general) and during the tether-release trials (specific). In young adults and older adults, COM displacement during the stepping phase significantly predicted landing phase COM displacement (Young: R^2=0.810, p<0.001; Older: R^2=0.746, p<0.001). Second, the specific models (but not general) were all significant predicators of COM displacement (p<0.001; R^2=0.569-0.783). The results of this study demonstrate the importance of specificity in predicting reactive stepping balance control. By emphasizing within-task metrics (i.e., step length or width) clinicians and researchers may be able to improve control during the landing phase of reactive stepping in older adults.

3.2 Introduction

The ability to effectively recover one's balance after a postural perturbation is of paramount importance, especially for preventing a fall. As such, many researchers have studied reactive stepping in response to a trip, slip etc., and how this ability changes with age. Older adults are slower to react and move with slower and shorter steps (Lee, et al., 2014; Luchies, et al., 1994; Schulz, et al., 2005; Thelen, et

al., 1997; Wojcik, et al., 1999). Regarding leg strength, conflicting reports exist. Researchers have suggested an important role for the hip, as older adult multi-steppers use a higher proportion of their hip extension strength compared to single-steppers (Carty, et al., 2012b). Similarly, others have implicated degeneration in leg-extensor muscle-tendon units as being an important factor in age-related differences during forward stepping (Karamanidis & Arampatzis, 2007; Karamanidis, et al., 2008). Oppositely, separate studies show that leg strength may play a minimal role in reactive stepping in older adults (Arampatzis, et al., 2008; Grabiner, et al., 2005). However, these studies did not focus on the landing phase of reactive stepping (after foot-contact).

Previous studies have revealed deficits during the landing phase in older adults (Singer, 2012; Singer, et al., 2013, 2016) (see section 2.5 in this thesis). Proper control of the center of mass (COM) when both feet are on the ground is very important, illustrated by the fact that incorrect weight shifting is a predominant cause of falls in frail older adults (Robinovitch, et al., 2013). Understanding if characteristics such as response time, movement amplitude or leg strength are predictive of COM movement during the landing phase will allow for the targeting of specific domains during interventions. Therefore, this study focused on the following objectives: 1) compare response time, movement amplitude, hip extension strength and COM displacement during reactive stepping, between young and older adults, to ensure that our between-group differences are similar to existing research; 2) determine if balance control during the stepping phase and landing phase is related, or if the characteristics present at the start of the landing phase represent a second perturbation, in young and older adults. During stepping, the support-leg can allow enough time and clearance for proper step-leg placement, and help to reduce angular momentum prior to the landing phase (Pijnappels, et al., 2004). However, it is unclear if body movement, or COM displacement, between phases is related. Further, research has suggested that both the COM displacement and velocity be considered when studying balance control, and when predicting the need to initiate a reactive step (Hof, et al., 2005; Pai, et al., 2000; Pai & Patton, 1997; Pai, et al., 1998); 3) assess the predictive value of a model consisting of general measures of response time, range-of-motion

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and leg strength, all measured separately from the tether-release trials; and 4) assess the predictive value of a model consisting of <u>specific</u> predictors drawn from the tether-release trials.

It was hypothesized that the older adults would have a delayed response time, reduced movement amplitude, lower hip extensions strength and larger COM displacement at and after foot-contact (FC) (objective 1), Regarding the second objective, it was hypothesized that COM displacement at FC would strongly (0.60-0.79) or very strongly (0.80-1.0) (Evans, 1996) correlate with peak COM displacement after FC, in both the young and older adults. Further, incorporating COM velocity at FC as a predictor would strengthen both models. Third, multiple linear regressions incorporating <u>general measures</u> would significantly predict COM displacement at FC (stepping phase) and peak COM displacement after FC (landing phase), in young and older adults, where the coefficient of multiple correlation (R) for each <u>general</u> statistical model would indicate moderate (0.40-0.59) to strong (0.60-0.79) (Evans, 1996) predictive ability for all models (objective 3). Fourth, the multiple-linear regressions incorporating <u>specific measures</u> would result in larger R values, vs. the general models, for each dependent variable and age-group (objective 4). For the purpose of this study, <u>general</u> refers to measures which were collected and calculated separately/in isolation from the balance recovery task (tether-release trials). Continuing with this framework, <u>specific</u> refers to measures calculated directly from the tether-release trials.

3.3   Methods

3.3.1   Experimental Protocol

Forty young adults (age: 22.7(3.3) y; height: 1.7(0.1) m; mass: 72.4(17.1) kg; 20 females) and 40 healthy, community-dwelling older adults (age: 69.6(4.3) y; height: 1.7(0.1) m; mass: 73.6(14.3) kg; 25 females) participated in the study. Ethics clearance was obtained from the University of Waterloo Human Research Ethics Committee prior to study commencement. Telephone interviews were conducted to ensure participant eligibility (Appendix 1). All participants provided their informed consent prior to participation.

Twelve cameras (Optotrak Certus, Northern Digital Incorporated, Waterloo, Ontario, Canada) were used to collect kinematic data at 64Hz, while three force-plates (BP 5050 (x2), Bertec, Columbus, Ohio, USA, and OR6-7 (x1), Advanced Mechanical Technology Inc., (AMTI) Watertown, MA, USA) were used to collect ground reaction forces and moments at 2048 Hz. The two Bertec force-plates were arranged side-by-side (Figures 3-1 and 3-2). The AMTI force-plate was where the participants stepped onto, using their right leg, during all stepping tasks. The tether supporting each participant's body weight was located in-line with a load cell (MLP-300-CO, Transducer Techniques, Temecula, CA) which was sampled at 2048 Hz and rated for up to136 kg. The tether was connected to a metal frame via an electromagnet (AEC Magnetics, Cincinnati, OH, USA).
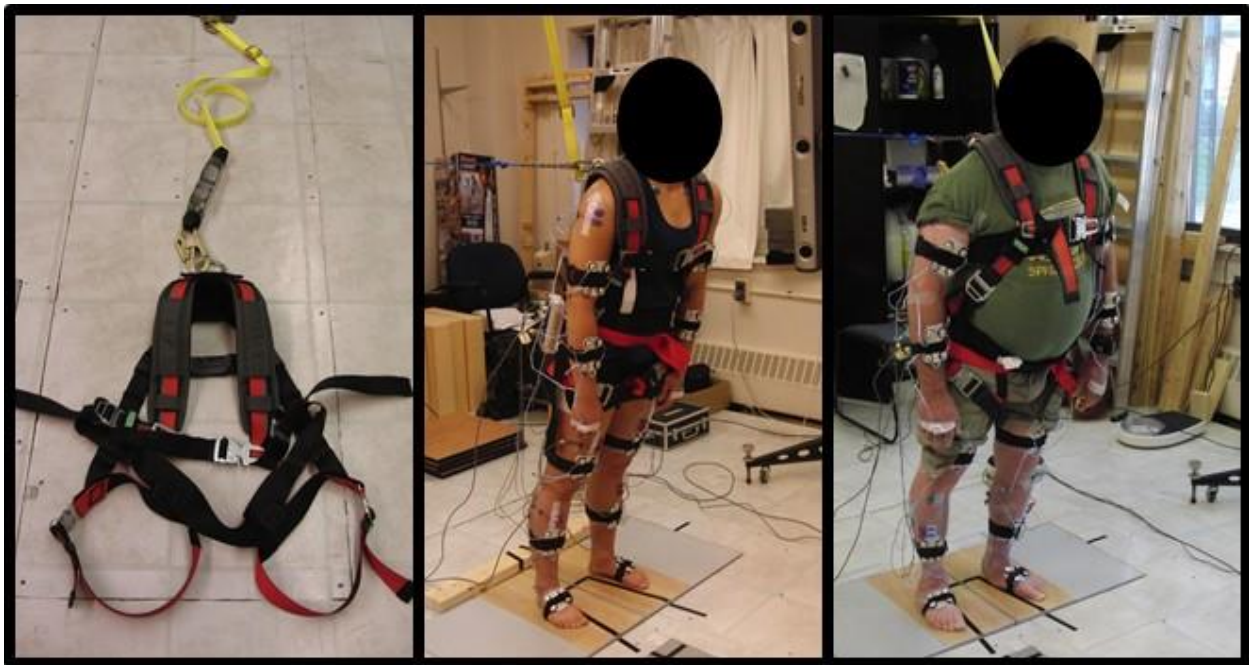
Figure 3-1: On the left is the safety harness which participants wore, along with the safety-tether which was connected to the ceiling to prevent participants from falling to the ground. In the middle, a young adult participant, and on the right, an older adult participant are depicted wearing the harness, along with the safety tether. The two Bertec force-plates are also depicted in both photos.
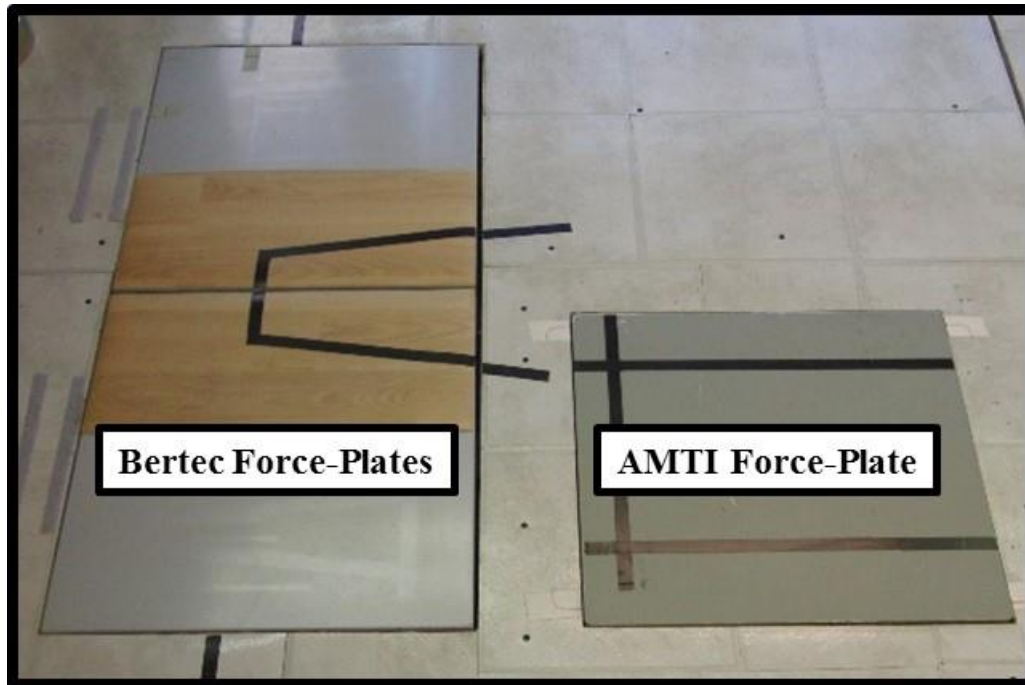
Figure 3-2: Depicted is the layout of the Bertec and AMTI force-plates. Marked with black electrical tape (on the Bertec force-plates) is the participants' initial foot position.

In the current study, ankle muscle electromyography (EMG) and full body kinematics were collected. Electromyographic data was sampled at 2048 Hz using a differential amplifier, with a hardware band-pass filter of 10-1000 Hz, a common mode rejection ratio of 115dB at 60 Hz (Bortec Biomedical, Calgary, AB) and disposable, self-adhesive Ag/Ag-Cl electrodes which were placed bilaterally on the tibialis anterior (TA) and medial gastrocnemius (MG). All data sources were synchronized using First Principles software (Northern Digital Incorporated, Waterloo, Ontario, Canada). The analog-to-digital converter included a 16-bit card. Kinematics were measured using a whole-body marker set. Rigid clusters of four markers were placed on the locations depicted in Appendix 2. Additionally, using a digitizing probe (Northern Digital Incorporated, Waterloo, Ontario, Canada) "imaginary" markers were digitized bilaterally at anatomically relevant locations (Appendix 2). The laboratory global coordinate system was defined in accordance with ISB recommendations (Wu & Cavanagh, 1995).

Hip extension strength was measured via a dominant-leg maximal isometric exertion against an ankle cuff, placed around the malleoli of the tibia (Figure 3-3). All participants completed the isometric

exertion twice. Foot-position was shoulder width apart and parallel. Participants were instructed to maintain an upright posture, minimize movement of their trunk, keep their knee straight, focus on movement at the hip and to keep the toes and heel in-line (i.e., avoid internal/external rotation). After ramping up to their maximum, each trial was collected for 3 seconds (Glinka, 2013).

Figure 3-3: A participant completing the isometric hip extension task. Note the arrow indicating the direction of pull, and the physiotherapy table for support. The participant did their best to keep their leg straight and focus on pulling with their hip muscles.

Maximal active range-of-motion (ROM) was assessed for hip flexion and hip abduction of the dominant leg. Participants completed five cycles of each movement within 20 seconds, and were instructed to maintain an upright posture, and avoid trunk movement as much as possible. During hip flexion, participants were told to keep their toes and heel in-line. During hip abduction, participants were told to lead with their heel. The goal of the task was to move their leg as far as possible in the specified direction, in a controlled manner, while avoiding forcing extra movement at the end (Figures 3-4 and 3-5). No movement speed was specified.

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Figure 3-4: A participant completing the active hip flexion range-of-motion task. Note the chair beside the participant which offered balance support if required.

Figure 3-5: A participant completing the active hip abduction range-of-motion task. Note the chair in front of the participant which offered balance support if required.

Voluntary reaction time was assessed by having participants start with their feet in a standardized position (0.17 m between heel centers and an angle of 14º between the long axes of the feet) (McIlroy & Maki, 1997), with one foot on each of the Bertec force-plates and their arms by their sides. Participants were instructed to focus their gaze to a computer monitor located at eye level, 3.28 m in front of them. In response to an auditory tone (presented randomly), participants stepped with their right leg as "fast as possible" onto the AMTI force-plate. A target was marked on the AMTI force-plate, which equalled a step length of 48 cm (Singer, et al., 2014) and a step width of 4 cm (McIlroy & Maki, 1996). Two trials were completed.

Next, two quiet standing trials were collected. The first had participants adopt the standardized foot position described above (McIlroy & Maki, 1997), with their arms at their sides, while looking straight ahead. The second trial required participants to start in the standardized foot position, take a single forward step onto the AMTI force-plate with their right leg, and hold the final forward-stance position (Singer, 2012). Both trials were 60 seconds in duration (Carpenter et al., 2001).

To provide the postural perturbations, a tether-release paradigm was used (Hsiao-Wecksler, 2008). Participants started in an initial forward-lean position pertaining to 10% of their body weight (Singer, et al., 2016), which was monitored in real-time and kept to within ± 1% body weight trial-to-trial (Graham, et al., 2015). Initial foot position was standardized as above (McIlroy & Maki, 1997). The center of pressure position of each foot and the body weight supported by each leg (Newtons) were monitored prior to each tether-release perturbation, using a real-time LabVIEW feedback routine (National Instruments Corporation, Austin, TX).

The EMG activity recorded bi-laterally from the TA and MG was used to help ensure the same levels of pre-perturbation activity prior to a tether-release. It was the goal that the level of pre-perturbation EMG activity observed during the tether-release trials would not exceed the maximum which was observed from participants during the feet side-by-side quiet standing trial (Singer, 2012; Singer, et al., 2016). This was done using a LabVIEW feedback routine (National Instruments Corporation, Austin, TX). Verbal encouragement was provided emphasizing participants to, "allow for the tether to fully

support their body weight". The tether was released at random intervals after the initial conditions were met. A minimum of one second was always captured from when the participant met the initial conditions, and when the electromagnet was powered off and the tether released.

Five practice trials were always completed first. Participants were instructed to respond to each tether-release trial by taking a single-step with their right leg. No additional restrictions were placed on the recovery step, as long as the entire foot landed on the AMTI force-plate. Participants started each trial with their arms at their sides, but no restriction was placed on arm movement after tether-release. Next, 10 preferred stepping trials were conducted in the same manner as the practice trials, where the goal was to respond with a single reactive step, using their right leg, onto the AMTI force-plate. Participants were instructed to maintain their final position for approximately 10 seconds once they regained their stability (Singer, 2012).

3.3.2   *Data Analysis*

All kinematic data (including the range-of motion trials) was low-pass filtered using a 2nd order, dual-pass, Butterworth filter with a cut-off frequency of 6 Hz (Graham, et al., 2015; Singer, et al., 2016). An estimate of the whole body COM was calculated using the filtered kinematic data and the anthropometric tables of de Leva (1996) for the young adults and Dempster (1955) (as displayed in Winter (2009)) for the older adults. To determine the hip and shoulder joint centers, the methods of Weinhandl and O'Connor (2010) and Nussbaum and Zhang (2000), respectively, were used. Next, the position of the COM, in the anterior-posterior (AP) and medio-lateral (ML) directions, was calculated at the following time points: 1) toe-off (TO), 2) FC and, 3) the peak COM position after FC. The peak COM after FC was chosen as a dependent variable, as it has previously been shown to be sensitive to differences between young and older adults during the landing phase (Singer, et al., 2013, 2016). Additionally, it represents the point of maximum body movement after FC, which assuming the foot-position does not change after FC, also represents the point of the minimum stability margin (after FC). Although many have advocated for accounting for both the position and velocity of the COM during dynamic tasks such as reactive stepping (e.g., via the extrapolated COM (xCOM)) (Hof, et al., 2005; Pai,

et al., 2000; Pai, et al., 1998), the COM was used to allow for direct comparison to recent studies focused on landing phase control (Singer, et al., 2016). In each instant, to calculate COM displacement, the COM was referenced to the mean starting COM value, which was calculated from the start of the trial (frame 1) to one frame before cable-release (CR). The peak ML COM position after FC was always calculated in the +Z (right) direction.

Force-plate data was also low-pass filtered using a 2nd order, dual-pass Butterworth filter with a cut-off frequency of 50 Hz (Singer, et al., 2016). Using the force-plates, TO and FC were defined. Toe-off was defined as the point when the vertical force under the right leg fell below 10 N (Sparrow & Tirosh, 2003), while FC was defined as the point when the vertical force signal of the force-plate (which participants stepped onto) exceeded, and remained above, 10 N (Sparrow & Tirosh, 2003). Cable-release was calculated using the data from the load cell in-series with the tether. This data was low-pass filtered using a 2nd order, dual-pass Butterworth filter, with a cut-off frequency of 3 Hz (Wright et al., 2014). Cable-release was defined, in accordance with previous research (Graham, et al., 2015), as a 20% reduction in force. The mean force value over the first second of the lean was used as the baseline value, which was then used to determine a CR threshold corresponding to a 20% reduction in tether force.

During the tether-release trials, reaction time was calculated as the time from CR to TO, while movement time was calculated from TO to FC. These two metrics were summed to calculate response time. An average was calculated over the 10 tether-release trials. From the stepping task (in response to the auditory tone), reaction time and movement time were also calculated using the same TO and FC thresholds described above. However, instead of starting at CR, reaction time was calculated from the time of auditory tone presentation to TO. From the two-stepping trials an average reaction time and movement time were calculated to determine response time.

Step length and width were calculated using the COM of the right foot. The difference in the position of the right foot COM between FC and CR was calculated as the step length (AP) and step width (ML), respectively. Both of these values were normalized to participant leg length (Graham, et al., 2015). An average was calculated over the 10 tether-release trials. Using the ROM trials, maximum active hip

flexion and hip abduction ROM were also calculated. Via the cosine law, these angles were calculated at the point of maximum hip flexion, and hip abduction. As the participants performed the ROM tasks in-place, these values were the maximum global anterior (hip flexion) and lateral (hip abduction) knee positions.

Maximum isometric hip extension strength was calculated by taking the maximum value during the last two seconds of each three second isometric hip extension trial (the three second recording started after the ramp period) (Viggiani, 2015). Each value was normalized to leg length and an average was taken using the two maximum values.

Lastly, the mean tether-load (during the lean), the mean vertical force under each foot and the mean start AP and ML COM positions were all calculated. Right foot AP and ML stability margins were calculated by subtracting the COM position from the tip of the big toe and the 5th metatarsal, respectively. All of these values were calculated from the start of the trial, to one frame prior to CR. Additionally, AP and ML COM displacements, stability margins, and velocities (central difference differentiation) were each determined at TO. COM displacements at TO were referenced to the mean starting COM value.

3.3.3   Statistical Analyses

To compare all variables between the young and older adults, independent samples t-tests were used (hypothesis 1). To determine if balance control between the stepping and landing phases was correlated, (hypothesis 2), linear-regressions were conducted with: 1) the COM displacement at FC, and 2) the COM displacement at FC and the COM velocity at FC as independent variables. Peak COM displacement after FC was the dependent variable. To assess hypothesis 3, for the COM displacement at FC, and the peak COM displacement after FC forced-entry multiple linear regressions were conducted incorporating general measures. The models included the following independent variables chosen apriori: 1) Response time (voluntary stepping task); 2) Maximum active hip flexion ROM; 3) Maximum active hip abduction ROM; 4) Normalized maximum isometric hip extension strength. To assess hypothesis 4, forced-entry multiple linear regressions were conducted incorporating specific measures of response time and movement amplitude (from the tether-release trials). Specifically, the models included the following

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variables selected apriori: 1) Response time (tether-release trials); 2) Normalized step length (Graham, et al., 2015); 3) Normalized step width; 4) Normalized maximum isometric hip extension strength (Table 3-1). All statistical analyses were conducted using SPSS (v.21, IBM Corporation, New York, USA). Experiment wide statistical significance was set at $p{\leq}0.05$.

Table 3-1: The rationale for why each predictor was included (#1-3), as well as why all predictor variables were included in a forced-entry multiple linear regression approach (#4).

1) <u>Response time</u> (composed of both reaction time and movement time) was selected due to the importance of a rapid step in response to a trip perturbation. Furthermore, older adults take reactive steps of a slower velocity after a forward or backward platform perturbation (Lee, et al., 2014). Both Thelen et al. (1997) and Wojcik et al. (1999) also noted that the older adults had slower reaction times compared to young adults.

2) Maximal active <u>hip flexion ROM</u> (with the knee straight) was selected because in order to generate a large step, one must be able to effectively flex their hip, to move the leg forward. Recent research has emphasized the importance of taking a long and rapid step after tripping (Graham, et al., 2015). Likewise, in order to modulate step width, one must have an adequate active <u>hip abduction ROM</u>. <u>Step length</u> and <u>width</u> were included (in the 'specific' model) due to previous studies showing age-related differences in the step length (Carty, et al., 2011; Cronin, et al., 2013; Graham, et al., 2014; Luchies, et al., 1994; Schulz, et al., 2005; Thelen, et al., 1997) and width (McIlroy & Maki, 1996; Rogers, et al., 2001; Schulz, et al., 2005; Troy, et al., 2008). Normalized step length was also implicated by Graham et al. (2015) as an important variable in predicting maximum recoverable lean angle. A medio-lateral component was included in the regressions (as opposed to step length alone) as recent research has quantified medio-lateral COM displacement after foot-contact, which is larger in older adults (Singer, et al., 2013, 2016).

3) <u>Hip extension strength</u> was chosen due to the observed greater trunk flexion in older adults during reactive stepping (compared to young adults) (Graham, et al., 2014). Similarly, Carty et al. (2011) found older adults to take shorter reactive steps and exhibit increased trunk flexion at FC compared to young adults. Both the trunk flexion angle at FC and step length correlated with the margin of safety at FC (Carty, et al., 2011). Crenshaw et al. (2012) observed that trunk flexion velocity at the first reactive step and trunk flexion angle at the second step had the most accurate overall classification of falls and recoveries, compared to step kinematics and stability measures. Further, older adult multi-steppers use a higher proportion of their hip extension strength (Carty, et al., 2012b). Interestingly, Madigan and Lloyd (2005a) also observed that during the support phase of a single reactive step older adults exhibited larger peak extensor torques at the hip and ankle. However, during gait termination older adults recruited fewer swing leg muscles with less frequent activation of the soleus and gluteus medius. Failure to activate muscles would provide less extensor torque, decreasing the total force opposing horizontal velocity (Tirosh & Sparrow, 2005).

4) The decision to include all predictor variables together was made due to the fact that mathematical models predict that successful balance recovery by stepping is governed by a coupling between step length, step execution time, and leg strength, so that the feasibility of balance recovery decreases unless declines in one capacity are offset by enhancements in the others, suggesting that one's risk for falls may be affected more by small but diffuse neuromuscular impairments than by larger impairment in a single motor capacity (Hsiao & Robinovitch, 1999).

In older adults, COM displacement at FC and the peak COM displacement after FC both tended

to be larger compared to the young adults (12.7 mm, $p=0.051$ and 20.2 mm, $p=0.053$, respectively). Of

the general predictors, only response time (i.e., in response to the auditory tone) was different between

age-groups (150 ms slower in older adults, $p<0.001$). For the specific predictors (i.e., in response to the

tether-release perturbations) older adults had a 60 ms slower response time ($p<0.001$), a 7.2% longer

normalized step length, and a 2.6% larger normalized step width. Older adults also had a 53.3 Nm smaller

normalized isometric hip extension strength ($p<0.001$) (Table 3-2). Mean(SD) values for the initial

conditions can be found in Table 3-3.

Table 3-2: Mean(SD) values for the dependent and independent variables used in the multiple linear regressions.

Variable	Young Adults	Older Adults	*p*-value
COM Disp. at FC (mm)	121.8(24.5)	134.5(32.0)	0.051
AP Disp.	119.6(24.5)	131.6(31.0)	0.059
ML Disp.	22.0(7.8)	26.6(11.3)	**0.036***
COM Velocity at FC (mm/s)	665.3(73.1)	643.4(81.3)	0.21
Peak COM Disp. after FC (mm)	243.6(42.2)	263.8(49.6)	0.053
AP Disp.	233.0(41.4)	249.8(46.4)	0.092
ML Disp.	69.2(17.8)	83.4(23.7)	**0.003***
Response Time (s)	0.80(0.13)	0.95(0.17)	**<0.001***
Max Hip Flexion ROM (deg)	83.8(19.9)	89.8(15.6)	0.138
Max Hip Abduction ROM (deg)	71.5(19.6)	70.3(18.3)	0.772
Response Time (tether-release) (s)	0.48(0.03)	0.54(0.08)	**<0.001***
Step Length (% Leg Length)	64.1(6.9)	71.3(8.5)	**<0.001***
Step Width (% Leg Length)	4.8(2.9)	7.4(3.7)	**<0.001***
Hip Ext. Strength (Nm)	161.8(49.7)	108.5(43.4)	**<0.001***

*indicates statistical significance at $p\leq0.05$ (independent samples t-test); COM = center of mass; FC = foot-contact; ROM = range of motion.

Table 3-3: Initial conditions for the young and older adults.

Initial Condition Variable	Young Adults	Older Adults	*p*-value
Tether Load (%BW)	10.4(0.6)	10.4(0.8)	0.99
Right Vertical Force (% Total)	50.0(0.4)	49.8(0.7)	0.32
Left Vertical Force (% Total)	50.0(0.4)	50.2(0.7)	0.32
AP COM Start Position (mm)	446.5(25.7)	448.7(37.8)	0.76
AP Stability Margin Before Release (mm)	-14.3(22.4)	-12.0(31.7)	0.71
AP COM Position at TO (wrt Start) (mm)	30.7(9.5)	35.5(12.6)	0.12
AP Stability Margin at TO (mm)	-40.4(22.3)	-41.0(33.1)	0.92
AP COM Velocity at TO (mm\s)	347.7(74.0)	341.5(78.9)	0.72
ML COM Start Position (mm)	8.8(4.2)	8.1(4.9)	0.47
ML Stability Margin Before Release (mm)	201.7(7.4)	206.5(9.2)	**0.01***
ML COM Position at TO (wrt Start) (mm)	1.7(2.4)	2.8(3.4)	0.09
ML Stability Margin at TO (mm)	192.9(7.1)	201.2(11.7)	**<0.001***
ML COM Velocity at TO (mm\s)	37.5(22.0)	49.2(24.9)	**0.03***

*indicates statistical significance at $p \leq 0.05$ (independent samples t-test)
BW = body weight; AP = anterior-posterior; ML = medio-lateral; COM = center of mass; TO = toe-off; wrt = with respect to.

Due to AP and ML correlations, the resultant COM displacements were used as the dependent variables (Table 3-4). In young adults and older adults, COM displacement at FC (stepping phase) was a significant predictor of peak COM displacement after FC (landing phase) (Young: R=0.900, R^2=0.810, p<0.001; Older: R=0.864, R^2=0.746, p<0.001). When COM velocity at FC was added as a predictor variable, the results for the young adults were very similar (R=0.901, R^2=0.812, p<0.001), and velocity was not a significant predictor (p=0.582). However, in the older adults, the addition of COM velocity at FC did account for 4% more explained variance (R=0.884; R^2=0.781, p<0.001), where displacement (p<0.001) and velocity (p=0.021) were both significant predictors in the model.

Table 3-4: Correlations between the COM displacement in the AP and ML directions, for young and older adults. Bold denotes $p \leq 0.05$.

	Young Adults		Older Adults	
	ML COM Disp. at FC	**ML Peak COM Disp. after FC**	**ML COM Disp. at FC**	**ML Peak COM Disp. after FC**
AP COM Disp. at FC	0.285; p=0.074	--------------------	0.693; **p<0.001**	--------------------
AP Peak COM Disp. after FC	--------------------	0.418; **p=0.007**	--------------------	0.684; **p<0.001**

AP = anterior-posterior; ML = medio-lateral; COM = center of mass; FC = foot-contact; Disp. = displacement.

In young adults, the combination of general predictor variables were not significant predictors of COM displacement at FC (R=0.192, F(4,35)=0.33, p=0.853) or the peak COM displacement after FC (R=0.135, F(4,35)=0.16, p=0.957). In the older adults, the same predictors resulted in a model which was a significant predictor of COM displacement at FC (R=0.635; R^2=0.403; F(4,35)=5.90, p=0.001), as well as the peak COM displacement after FC (R=0.483; R^2=0.233; F(4,35)=2.66, p=0.049). In both older adult models, hip-extension strength was the only significant variable (Table 3-5).

Table 3-5: Regression coefficients, standard error, and individual factor significance values for the models with general response time and range-of-motion variables.

	Factor	Young Adults				Older Adults			
		B	**SE**	**t**	**Sig.**	**B**	**SE**	**t**	**Sig.**
FC	Response Time	4.20	35.92	0.12	0.91	-1.29	25.59	-0.05	0.96
	Hip Flexion	0.20	0.27	0.77	0.45	-0.01	0.40	-0.03	0.98
	Hip Abduction	0.04	0.27	0.14	0.89	0.11	0.33	0.33	0.74
	Hip Ext. Strength	-0.01	0.10	-0.09	0.93	-0.46	0.10	-4.66	**<0.001**
Peak after FC	Response Time	33.17	62.30	0.53	0.60	23.84	44.94	0.53	0.60
	Hip Flexion	0.26	0.46	0.56	0.58	-0.04	0.70	-0.05	0.96
	Hip Abduction	-0.08	0.46	-0.17	0.87	0.09	0.58	0.15	0.88
	Hip Ext. Strength	0.06	0.17	0.35	0.73	-0.55	0.17	-3.15	**0.003**

Hip flexion and hip abduction refer the active range-of-motion tasks collected before the tether-release trials. FC = foot-contact; B = beta; SE = standard error.

For both age groups, trial specific variables were significant predictors of COM displacement. For the young adults, 72% of the variance in COM displacement at FC was explained by the model ($R=0.848$, $F(4,35)=22.42$, $p<0.001$), while the combination of predictors explained 64% of the variance in the peak COM displacement after FC ($R=0.799$, $F(4,35)=15.50$, $p<0.001$). Response time was significant in both models for the young adults, while step length was significant only for the COM displacement at FC (Table 3-6). In older adults, 78% of the variance in the COM displacement at FC was explained ($R=0.885$, $F(4,35)=31.54$, $p<0.001$), while 57% of the variance in the peak COM displacement after FC was accounted for ($R=0.754$, $F(4,35)=11.56$, $p<0.001$). In both models, normalized step length was the only significant predictor for the older adults (Table 3-6). Scatter-plots of the actual vs. predicted COM displacement from the specific models are depicted in Figure 3-6.

Table 3-6: Regression coefficients, standard error, and individual factor significance values for the models with specific response time and movement amplitude variables.

	Factor	Young Adults				Older Adults			
		B	**SE**	**t**	**Sig.**	**B**	**SE**	**t**	**Sig.**
FC	Response Time	452.71	82.40	5.49	**<0.001**	44.66	34.73	1.29	0.21
	Step Length	1.23	0.46	2.66	**0.01**	2.75	0.39	7.04	**<0.001**
	Step Width	-0.34	0.91	-0.37	0.71	0.39	0.80	0.49	0.63
	Hip Ext. Strength	0.05	0.05	1.13	0.27	-0.09	0.08	-1.18	0.25
Peak after FC	Response Time	868.42	160.45	5.41	**<0.001**	-3.49	75.79	-0.05	0.96
	Step Length	1.24	0.90	1.38	0.18	3.86	0.85	4.53	**<0.001**
	Step Width	-1.09	1.76	-0.62	0.54	2.14	1.75	1.22	0.23
	Hip Ext. Strength	0.12	0.09	1.37	0.18	-0.01	0.17	-0.06	0.95

FC = foot-contact; B = beta; SE = standard error.
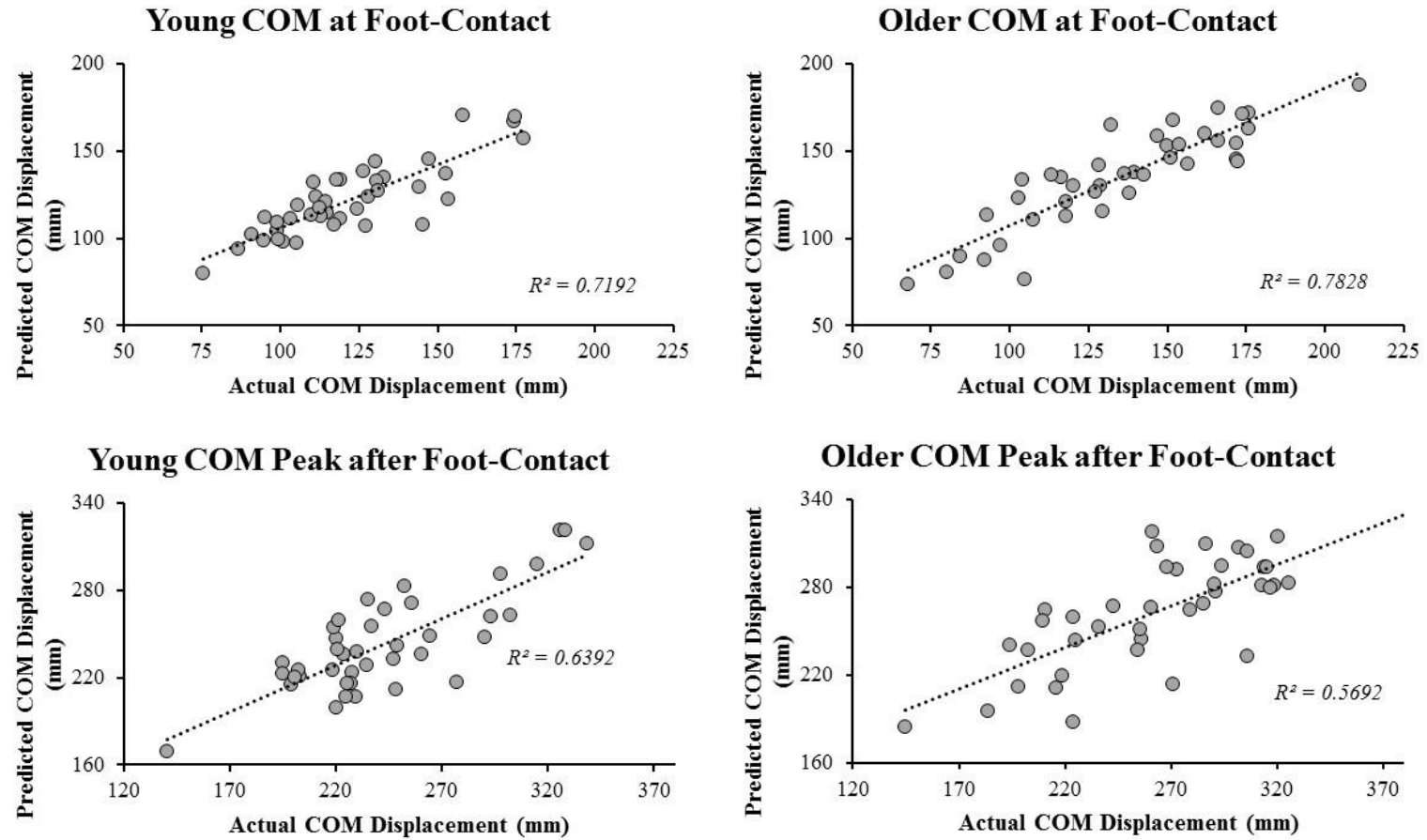
Figure 3-6: Scatter-plots of the actual vs. predicted center of mass (COM) displacement at foot-contact (top row), and the peak COM displacement which occurred after foot-contact (bottom row), in young (left column) and older (right column) adults.

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3.5   Discussion

The primary purposes of this study were to compare individual characteristics and COM displacement during reactive stepping between young and older adults, determine if COM displacement during the stepping phase was related to that during the landing phase, and to determine if general participant characteristics, and trial specific responses, were predictive of COM displacement at different points during reactive stepping. In agreement with our first hypothesis, older adults exhibited larger COM displacements both at and after FC. Second, COM displacement during the stepping and landing phase were very strongly correlated (.80-1.0) (Evans, 1996) in both age-groups (Young: r=0.900; Older: r=0.864), while COM velocity at FC was also a significant predictor in older adults. However, while the general measures did have moderate (.40-59) to strong (.60-.79) (Evans, 1996) predictive utility in the older adults, the specific models performed superiorly for predicting COM displacement during both phases. As hypothesized, this was true for young and older adults, highlighting the importance of task specificity in predicting balance control during reactive stepping.

As hypothesized older adults had generally slower response times, longer and wider steps, reduced hip extension strength, and tended to exhibit increased COM displacement at both time-points. Our observation of significant associations between COM displacement (and velocity) during the stepping and landing phases of reactive stepping also builds upon previous literature. In line with our second hypothesis, COM displacement at FC was a strong predictor of peak COM displacement after FC, for both young and older adults. In the older adults, (but not young adults) there was an increase in explained variance when COM velocity at FC was added. Using the extrapolated COM proposed by Hof et al. (2005), Carty et al. (2011) observed the margin of safety (MoS) at FC to be significantly correlated (r=0.88) with MoS at the maximum knee flexion angle after FC, however the latter time point does not necessarily correspond to the maximum COM displacement after FC. The current study expands this finding to encompass the entire landing phase, suggesting that COM movement after FC during single-step responses may be due to poor-control during the stepping phase. Interestingly, during lateral-directed stepping in older adults, multiple-step stability margins (at higher intensity perturbations) were smaller at

first step lift-off, compared to single-steps. This suggests that multiple steps could be attributable to COM dynamics as early as first step lift-off (Fujimoto et al., 2017).

The third and fourth objectives of this study were to assess the predictive value of models consisting of <u>general</u>, as well as <u>specific</u> measures, for predicting COM displacement during the stepping and landing phases. Overall, we observed stronger regressions with specific predictor variables, which was in-line with our final hypothesis. Recent research in older adults found that task-specific variables including normalized step length and peak hip extension moment during stepping together accounted for 69% of the variance in the maximum recoverable lean magnitude (Graham, et al., 2015). In the current study, the combination of specific predictor variables predicted COM displacement at FC with 78% explained variance, and the peak displacement after FC, with 57% explained variance, in older adults. However, our dependent variable was COM displacement, calculated from specific time points in the time-varying COM waveform, while Graham et al. (2015) focused on the maximum recoverable lean angle. During gait termination, older adults frequently use two-step responses when within their predicted stability region prior to taking the second step, suggesting that the second step was not needed (Tirosh & Sparrow, 2004). This result shows the importance of including a mechanical metric of balance control, as done in the current study, because older adults may take a second step when it is not required mechanically. Further, the results of Graham et al. (2015) provide no insight into balance control at different points in time (of the response), and whether their predictor variables relate to control during separate phases of the reactive stepping response. The results of the current study add to the literature by showing that an objective, quantitative measure of balance control, (COM displacement), can be predicted using specific measures calculated from the trial itself.

In the specific models, normalized step length was positively associated with COM displacement at both time-points for the older adults. However, in the young adults, response time was a significant, positive predictor in both specific models. As the older adults had a 60 ms slower response time during the tether-release trials, this suggests a difficulty in moving as quickly as the young adults. For older adults step length is important because a delayed response time could cause the COM to be displaced

further forward prior to the moment of FC (statistical trend of p=0.059, Table 3-2). Specifically, the COM of the older adults was displaced 11.98 mm further in the anterior direction at FC, compared to the young adults. Therefore, an increased step length would be required to adequately capture the falling COM. In the current study, older adults were on average 60 ms slower in their response time compared to young adults (Table 3-2), which is very close to the 72 ms mean difference reported by Wojcik et al. (1999). The positive associations between step length and COM displacement, and the fact that COM displacement was correlated between phases ($r > 0.85$) suggests that COM displacement after FC in older adults may manifest due to poor control during earlier phases of the stepping reaction (where a large step length is needed to account for the larger COM displacement), as opposed to an active strategy (Singer, et al., 2013). Interestingly, the general models did result in significant predictions of COM displacement in the older adults, but not the young adults. For both measures of COM displacement, normalized hip extension strength was the only significant predictor variable. Perhaps the young adults did not require maximal levels of hip extension strength, as their mean value was 53.28 Nm greater compared to the older adults. Previous research suggests that low-strength older adults are at a greater risk of falling during after-step falls (471 to 785 ms after recovery foot ground contact) (Pavol, et al., 2001, 2002). After-step falls have been suggested to be related to lower-extremity weakness, primarily of the hip and knee extensors, as primary factors in such falls were excessive lumbar flexion and buckling of the recovery limb (Pavol, et al., 2001, 2002). Our results indirectly support this notion as hip extension strength was negatively associated with COM displacement in both older adult regressions.

This study was associated with several limitations. First, the fact that participants were instructed to recover their balance using a single step, coupled with the uni-directional perturbation, likely promoted more voluntary, or pre-planned components to the reactive stepping response. Accordingly, multi-directional paradigms such as waist-pulls or surface translations may be better suited for evoking reactive steps with minimal pre-planning. Therefore, while the results of the current study do carry implications for training reactive stepping, they must be weighed in accordance with this limitation. Note, this initial limitation is relevant to all studies presented in this thesis. Second, the stance configuration used in the

active ROM trials required the participants to exert substantive muscle moments (especially at the end ROM). As such, it is unclear whether end ROM may have also required certain levels of strength to move the leg against gravity. Perhaps the older adults employed compensatory techniques, which may be why the young and older adults did not differ in ROM values (Table 3-2). Third, the isometric exertion may have been unfamiliar to many participants, possibly influencing their ability to perform the task. However, multiple trials were collected in an attempt to minimize this possibility. Fourth, the perturbation used was highly controlled, and does not truly mimic a real-life fall. Researchers may address this concern (to a degree) by using only the first (practice) trial to calculate COM displacement in follow-up studies. Fifth, the older adults were community-dwelling, and not likely at a high risk of falling. Although community-dwelling older adults, on average, are at a lower risk of falls, it remains possible that they were fearful or nervous of completing the tether-release task. To limit the influence of psychological factors in reactive stepping analysis, future studies may wish to include older adults who exhibit high activity levels, with minimal fear of falling, etc. Such a sample would allow researchers to more closely isolate the physiological effects of aging alone, without concomitant psychological factors that may be present in older adults. Sixth, during the reaction time task, despite verbal encouragement, there was no way to guarantee participants were stepping as fast as possible. More exhaustive sensory tests should be incorporated moving forward. Lastly, a larger sample size would allow for response time to be divided into reaction and movement time components in the regression models. However, both reaction time and movement time (during the tether-release trials) were delayed in the older, compared to young adults (similar to response time). Limitations aside, this study was novel as it focused on prediction during the landing phase (beyond FC).

In conclusion, COM displacement during the stepping and landing phase was correlated in both age-groups. Second, the specific regressions performed better than general models for predicting COM displacement. By emphasizing within-task metrics (such as response time, step length or width), clinicians and researchers may be able to improve landing phase control in older adults. However, caution must be exercised, given the predictable nature of the tether-release paradigm, and the instructional set

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employed, where participants were told to respond with a step. Next steps must focus on older adults who are at an elevated risk of falling, as well as individuals with neurological conditions. Future studies (Chapter 4) should also characterize lower-limb muscle recruitment, as this information could be an important part of developing interventions to train reactive stepping responses.

4. STUDY TWO – CHARACTERIZING PEAK LOWER-LIMB MUSCLE ELECTROMYOGRAPHY DURING THE LANDING PHASE OF REACTIVE STEPPING

4.1 Chapter Overview

Many studies have reported that during balance recovery tasks, older adults demonstrate impaired muscle responses. However, lower-limb muscle recruitment has not been characterized during the landing phase of reactive stepping. Accordingly, the primary objective of this study was to quantify lower-limb muscle peak recruitment patterns during this important element of dynamic balance control. A secondary objective was to assess between-muscle relationships in peak magnitude within each phase. Twenty young adults participated in this study. A tether-release paradigm was used to evoke 10 reactive stepping responses. Electromyography (EMG) was recorded bilaterally from the rectus femoris, biceps femoris, tibialis anterior and medial gastrocnemius. Peak timing and magnitude were generally slowest (most variable) and smallest from the peak center of mass after foot-contact to trial end. The muscles which exhibited their highest peak magnitude during the landing phase were the biceps femoris of the stepping leg, and the rectus femoris and tibialis anterior of the support-leg. Additionally, the biceps femoris and medial gastrocnemius peak magnitudes were significantly correlated during the landing phase, in both legs. High peak magnitudes of the step-leg bicep femoris and medial gastrocnemius, along with the significant correlations, suggest the step-leg extensors are important during foot landing, while the step-leg rectus femoris and tibialis anterior may be more crucial during the swing phase. This study provided a basic science characterization of leg muscle patterns in young adults. Clinicians can target phases of reactive stepping by strengthening the muscles which were most active in a given phase.

4.2 Introduction

Numerous studies have examined muscle recruitment and balance recovery in older adults. In response to tether-release perturbations, older adult multi-step responders showed reduced peak electromyography (EMG) in 6 of 7 step-leg muscles (Cronin, et al., 2013). However, researchers focused

only on the stepping phase up to foot-contact (FC), making it unclear how muscle recruitment differs (if at all) during the landing phase of reactive stepping. Older adults also exhibit delayed EMG responses in all postural leg, hip, trunk and arm muscles after rotational perturbations (Allum, et al., 2002), while anterior-posterior surface translations revealed delayed onset latencies in the medial gastrocnemius and biceps femoris in older adults (Tokuno, et al., 2010). Older adults also recruit fewer swing leg muscles during gait termination (Tirosh & Sparrow, 2005).

While these studies showed differential muscle recruitment in older adults, lower-limb muscle recruitment during the landing phase of reactive stepping has yet to be quantified. Balance control during this phase of reactive stepping has important implications for avoiding a fall after a perturbation, where age-related changes have previously been reported in older adults (Singer, et al., 2016). As such, knowledge of the muscle control strategies could assist with the development of appropriately targeted exercise interventions. Therefore, the primary objective of this study was to assess baseline lower-limb muscle recruitment patterns (peak timing and magnitude) over the entirety of the reactive stepping response, including after the peak center of mass (COM) position following FC, which is typically the kinematic event of interest in analysis of the landing phase (Singer, et al., 2013, 2016). Means and the coefficient of variation were assessed to gain insight into trial-to-trial variability, in-line with previous research which has reported on trial-to-trial variability during the landing phase (Singer, et al., 2012, 2013, 2016). Additionally, if a muscle (in a specific phase) exhibits low variability, along with a large peak magnitude, this may reflect a stereotypical response suggesting that the muscle is crucial for balance control in a given phase. A secondary objective was to assess between-muscle relationships in peak magnitude within each phase, in an attempt to gain insight into synergistic activity between muscles.

It was hypothesized that: 1a) step-leg rectus femoris peak timing and magnitude would be earliest and largest during toe-off (TO) to FC; 1b) step-leg biceps femoris peak timing and magnitude would be earliest and largest during FC to the peak COM position after FC, while the support-leg biceps femoris activity would be earliest and largest from cable-release to TO; 1c) step-leg tibialis anterior peak timing and magnitude would be earliest and largest from TO to FC; 1d) medial gastrocnemius (of both legs) peak

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timing and magnitude would be earliest and largest from cable-release to TO. The peak timing and magnitude of the medial gastrocnemius of both legs would also be early and large during FC to the peak COM position after FC, to slow the falling COM along with the biceps femoris as hypothesized above. Overall, for each muscle, the period from the peak COM after FC to trial end would have the latest peak timing and smallest peak magnitude, with the largest variability. 2a) During cable-release to TO, all four muscles (of both legs) would positively correlate; 2b) During TO to FC, the rectus femoris and tibialis anterior of the stepping leg would positively correlate, as the leg swings forward; 2c) During FC to the peak COM position after FC, the biceps femoris and medial gastrocnemius of the stepping leg would exhibit the largest positive correlation. Previous research has shown that the triceps surae and hamstring are important for restraining forward rotation and generating a push-off force in the support limb (Pijnappels, et al., 2004; Pijnappels et al., 2005; Pijnappels et al., 2008). It is proposed that the stepping leg biceps femoris and medial gastrocnemius may operate in a similar manner during step landing; 2d) During the peak COM position after FC to trial end, the peak magnitudes of the i) rectus femoris and medial gastrocnemius and ii) biceps femoris and tibialis anterior would each positively correlate during this 'quasi'-static phase (i.e., flexors and extensors working together to control sway).

4.3   Methods

4.3.1   Experimental Protocol

A subset of 20 young adults participated in Study 2 (10 males and females, mean(SD) age = 22.4(3.1) y; height = 1.7(0.1) m; mass = 74.0 (20.6) kg; participants were a subset of a larger collection described in Chapter 3, section 3.3.1). Ethics clearance was obtained from the University of Waterloo Human Research Ethics Committee. All participants provided informed consent prior to participation.

Twelve cameras (Optotrak Certus, Northern Digital Incorporated, Waterloo, Ontario, Canada) were used to collect kinematic data at 64Hz, while three force-plates (BP 5050 (x2), Bertec, Columbus, Ohio, USA, and OR6-7 (x1), Advanced Mechanical Technology Inc., (AMTI) Watertown, MA, USA) were sampled at 2048 Hz. The two Bertec force-plates were arranged side-by-side (Figure 4-1 and Figure 4-2) under the initial stance limbs. The AMTI force-plate was positioned to the right side, anterior to the

Bertec force-plate, to capture the entire landing phase of the step response (stepping was always performed with the right leg). The tether supporting each participant's body weight was located in-line with a load cell (MLP-300-CO, Transducer Techniques, Temecula, CA) which was sampled at 2048 Hz and rated for up to 136 kg. The tether was connected to a metal frame via an electromagnet (AEC Magnetics, Cincinnati, OH, USA).
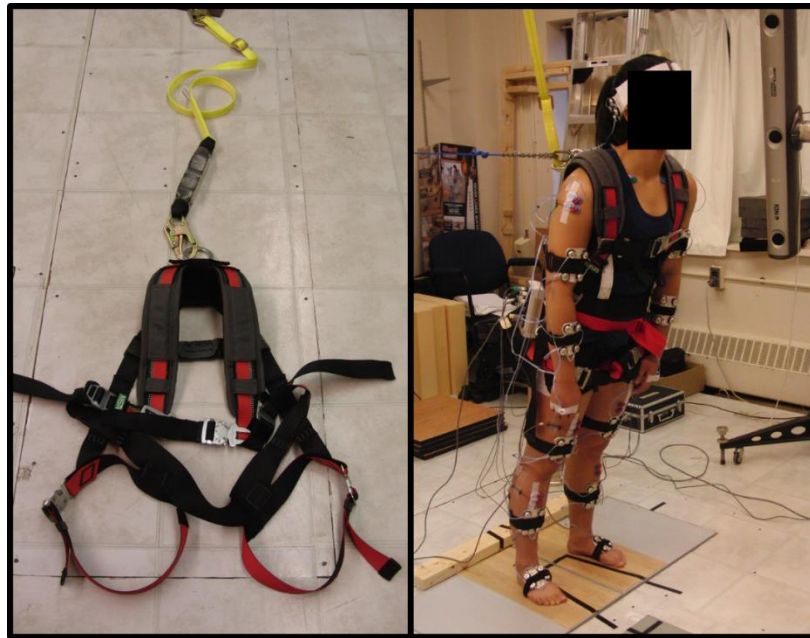
Figure 4-1: On the left is the safety harness which participants wore, along with the safety-tether which was connected to the ceiling to prevent participants from falling to the ground. On the right, a participant is depicted wearing the harness, along with the safety tether. The two Bertec force-plates are also depicted.
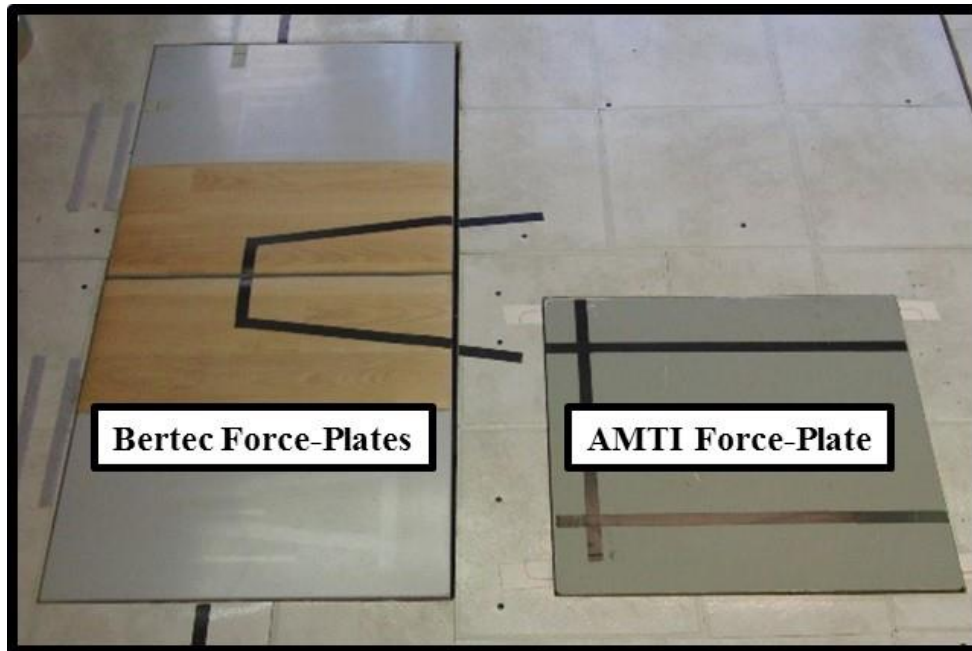
Figure 4-2: Depicted is the layout of the Bertec and AMTI force-plates. Marked with black electrical tape (on the Bertec force-plates) is the participants' initial foot position.

In the current study, lower-limb EMG and full body kinematics were collected. Electromyographic data was sampled at 2048 Hz using a differential amplifier, with a hardware band-pass filter of 10-1000 Hz, a common mode rejection ratio of 115dB at 60 Hz (Bortec Biomedical, Calgary, AB) and disposable, self-adhesive Ag/Ag-Cl electrodes which were placed bilaterally on the rectus femoris (RF), biceps femoris (BF), tibialis anterior (TA) and medial gastrocnemius (MG). All motion capture data and analog-to-digital converted signals were synchronized using First Principles software (Northern Digital Incorporated, Waterloo, Ontario, Canada). The analog-to-digital converter included a 16-bit card. Kinematics were measured using a whole-body marker set. Rigid clusters of four markers were placed on the locations depicted in Appendix 2. Additionally, using a digitizing probe (Northern Digital Incorporated, Waterloo, Ontario, Canada) "imaginary" markers were digitized bilaterally at anatomically relevant locations (Appendix 2). The laboratory global coordinate system was defined in accordance with ISB recommendations (Wu & Cavanagh, 1995).

The experimental protocol consisted of the participant completing five (5) practice tether-release trials, followed by 10 preferred stepping tether-release trials. The details associated with the tether-release protocol were presented earlier in the document (see details in Chapter 3, section 3.3.1).

4.3.2 Data Analysis

All eight channels of EMG were linear enveloped by subtracting the mean bias, full-wave rectifying and low-pass filtering each signal using a 2nd order, single-pass, Butterworth filter with a cut-off frequency of 3 Hz (Brenneman, 2014; Winter, 2009; Winter & Yack, 1987). This cut-off corresponds to twitch-response times reported by previous authors (Milner-Brown et al., 1973a, 1973b). Linear enveloping helps to produce a signal that mimics the twitch response of a muscle, and in a graded contraction, mimics the superposition of muscle twitches, by (attempting) to account for electromechanical delay. Recall that filter cut-off (F_c) is mathematically related to a given muscle's twitch time (T) via the following formula for a critically damped second-order low-pass filer: $F_c = 1/2\pi T$. This results in a signal that resembles the timing and shape of a muscle's tension curve (Winter, 2009). Further, the raw EMG signals were low-pass filtered in an attempt to remove the random high frequency noise inherent to EMG signals that might influence a reliable detection of peak magnitude (O'Connell et al., 2016). To normalize each participant's EMG signals, the peak of the averaged signal (for each muscle) was used (Yang & Winter, 1984). This method has previously been shown to reduce inter-subject variability compared to normalizing to 50% maximum voluntary contraction (Yang & Winter, 1984), and continues to be employed in current biomechanical research (Harper et al., 2014; Lockhart & Kim, 2006; Nüesch et al., 2016).

For all analyses, cable-release was calculated using the data from the load cell located in-series with the tether. This data was low-pass filtered using a 2nd order, dual-pass Butterworth filter, with a cut-off frequency of 3 Hz (Wright, et al., 2014). The point at which cable-release occurred was defined in accordance with previous research (Graham, et al., 2015). Specifically, cable-release was defined as a 20% reduction in force measured using the load cell located in-series with the tether attached to the participant's harness. All kinetic data collected from the force-plates was low-pass filtered using a 2nd

order, dual-pass, Butterworth filter with a cut-off frequency of 50 Hz (Singer, et al., 2016). Toe-off was defined as the time point when the vertical force under the right leg fell below 10 N (Sparrow & Tirosh, 2003), while FC was defined as the time point when the vertical force signal of the force-plate (which participants stepped onto) exceeded, and remained above, 10 N (Sparrow & Tirosh, 2003). Next, from each normalized EMG signal, the peak timing and peak magnitudes were calculated for each trial during the following phases: Cable-release to TO (phase1); TO to FC (phase 2); FC to the peak COM after FC (phase 3); the peak COM after FC to trial end (phase 4). Mean peak timing and magnitude values were calculated for each participant, for each muscle, during each of the four phases. Peak timing was calculated as the time between the event of interest (i.e., Cable-release, TO, FC, etc.) and the peak EMG magnitude, similar to previous research (O'Connell, et al., 2016), which assessed peak EMG timing and magnitude during slips. Peak timing has been used previously to assess potential adaptions to perturbation based training in older adults (Parijat et al., 2015), as well as to distinguish muscle activation patterns during step recovery in older adult women with and without a history of falls (Ochi et al., 2014). For each participant, muscle and phase, the mean coefficient of variation (CV) was also calculated for the peak timing and magnitude to gain insight into trial-to-trial variability between phases for each muscle.

4.3.3   *Statistical Analyses*

To compare the peak timing and magnitude mean and CV values (hypotheses 1a – 1d) Friedman tests were used with phase (four phases; as defined above) as the within-subjects factor. Post-hoc Wilcoxon Signed-Rank tests were used when appropriate to examine main effects of phase. These non-parametric tests were used as the EMG variables were not normally distributed, as assessed using Shapiro-Wilk tests, where $p \leq 0.05$. To assess the relationship between each muscle's peak magnitude within each phase (hypotheses 2a – 2d), Spearman Rank-Order correlations were used. For all correlations, correlation strength was interpreted according to guidelines suggested by Evans (1996): moderate (.40–59); strong (.60–.79); and very strong (.80–1.0). All statistical analyses were conducted using SPSS (v.21, IBM Corporation, New York, USA). Experiment wide statistical significance was set at $p \leq 0.05$.

4.4   Results

In all 16 peak timing analyses, a main effect of phase was observed for each muscle

($X^2(3)$=28.02–54.85; p<0.001). For each muscle, peak timing during phase 4 (the peak COM after FC to

trial end) was the slowest, except when compared to phase 1 for both the step-leg RF and the TA (Table

4-1). Step-leg peak timing mean(SD) values ranged from: <u>RF</u>: 0.057(0.016) *(phase 2)* to 0.851(0.969)

(phase 4) seconds; <u>BF</u>: 0.128(0.051) *(phase 2)* to 1.74(1.91) *(phase 4)* seconds; <u>TA</u>: 0.036(0.016) *(phase*

2) to 0.667(0.813) *(phase 4)* seconds; <u>MG</u>: 0.052(0.032) *(phase 3)* to 1.44(1.31) *(phase 4)* seconds.

Support-leg peak timing mean(SD) values ranged from: <u>RF</u>: 0.052(0.039) *(phase 3)* to 1.15(0.99) *(phase*

4) seconds; <u>BF</u>: 0.001(0.002) *(phase 2)* to 2.04(1.82) *(phase 4)* seconds; <u>TA</u>: 0.110(0.027) *(phase 3)* to

1.17(1.22) *(phase 4)* seconds; <u>MG</u>: 0.006(0.011) *(phase 2)* to 2.47(1.78) *(phase 4)* seconds.

Table 4-1: Mean(SD) for the peak timing mean values for each muscle and phase. The timing values
reported are relative to the start of each phase, and are in seconds.

Muscle		Phase 1	Ph. 2	Ph. 3	Ph. 4	Friedman Test
R	RF	0.303(0.025) [2,3]	0.057(0.016) [1,4]	0.079(0.067) [1,4]	0.851(0.969) [2,3]	$X^2(3)$=32.64; ***p<0.001***
	BF	0.215(0.018) [2,3,4]	0.128(0.051) [1,3,4]	0.144(0.292) [1,2,4]	1.74(1.91) [1,2,3]	$X^2(3)$=28.02; ***p<0.001***
	TA	0.280(0.039) [2]	0.036(0.016) [1,3,4]	0.255(0.099) [2,4]	0.667(0.813) [2,3]	$X^2(3)$=38.88; ***p<0.001***
	MG	0.228(0.018) [2,3,4]	0.089(0.063) [1,4]	0.052(0.032) [1,4]	1.44(1.31) [1,2,3]	$X^2(3)$=51.48; ***p<0.001***
L	RF	0.267(0.037) [2,3,4]	0.150(0.027) [1,3,4]	0.052(0.039) [1,2,4]	1.15(0.99) [1,2,3]	$X^2(3)$=51.87; ***p<0.001***
	BF	0.251(0.028) [2,3,4]	0.001(0.002) [1,3,4]	0.170(0.052) [1,2,4]	2.04(1.82) [1,2,3]	$X^2(3)$=49.80; ***p<0.001***
	TA	0.267(0.025) [2,3,4]	0.129(0.052) [1,4]	0.110(0.027) [1,4]	1.17(1.22) [1,2,3]	$X^2(3)$=38.34; ***p<0.001***
	MG	0.264(0.026) [2,3,4]	0.006(0.011) [1,3,4]	0.233(0.284) [1,2,4]	2.47(1.78) [1,2,3]	$X^2(3)$=53.46; ***p<0.001***

R = right; L = left; RF = rectus femoris; BF = biceps femoris; TA = tibialis anterior; MG = medial
gastrocnemius; Ph. = phase. Superscript values refer to the phases which statistically differ from the given
phase.

For the peak timing variability, as reflected by mean CV values, a main effect of phase was

observed for each muscle, in all 16 analyses ($X^2(3)$=24.47–44.10; p<0.001). In 50% of the muscles

studied, phase 4 variability was larger than that of phases 1,2 and 3 (step-leg RF and TA; support-leg BF and TA) (Table 4-2).

Table 4-2: Mean(SD) for the peak timing coefficient of variation values for each muscle and phase.

Muscle		Phase 1	Ph. 2	Ph. 3	Ph. 4	Friedman Test
R	RF	6.5(4.4) [2,3,4]	25.5(16.4) [1,4]	81.5(97.7) [1,4]	160.7(47.5) [1,2,3]	$X^2(3)=39.18$; *p<0.001*
	BF	7.5(3.1) [2,3,4]	49.2(62.1) [1,3,4]	89.8(61.1) [1,2]	137.3(69.8) [1,2]	$X^2(3)=37.08$; *p<0.001*
	TA	9.8(5.9) [2,3,4]	51.6(44.0) [1,4]	66.8(38.9) [1,4]	138.2(61.9) [1,2,3]	$X^2(3)=44.10$; *p<0.001*
	MG	7.7(3.1) [2,3,4]	102.3(99.2) [1]	50.9(28.3) [1,4]	132.4(48.0) [1,3]	$X^2(3)=34.26$; *p<0.001*
L	RF	18.5(22.2) [3,4]	17.8(17.6) [3,4]	112.1(39.7) [1,2]	139.6(66.2) [1,2]	$X^2(3)=39.35$; *p<0.001*
	BF	8.4(4.0) [2,3,4]	54.2(69.7) [1,4]	39.6(23.5) [1,4]	147.7(68.9) [1,2,3]	$X^2(3)=31.35$; *p<0.001*
	TA	9.2(2.8) [2,3,4]	61.5(75.3) [1,4]	23.2(14.0) [1,4]	137.0(52.4) [1,2,3]	$X^2(3)=43.92$; *p<0.001*
	MG	11.1(5.7) [2,3,4]	112.5(93.8) [1]	134.7(102.3) [1]	114.7(48.5) [1]	$X^2(3)=24.47$; *p<0.001*

R = right; L = left; RF = rectus femoris; BF = biceps femoris; TA = tibialis anterior; MG = medial gastrocnemius; Ph. = phase. Superscript values refer to the phases which statistically differ from the given phase.

Regarding the peak magnitude, a main effect of phase was also observed for each muscle ($X^2(3)=10.38$–53.46; $p=0.016$ – $p<0.001$). In the stepping leg, for each muscle, phase 4 was the smallest compared to all other phases, except for phase 1 for the TA. The muscles which exhibited their highest peak magnitude during the landing phase (either phase 3 or phase 4) were the BF of the stepping leg (98.4(23.9) % peak activity), and the RF (110.2(14.4) % peak activity) and TA (111.0(12.3) % peak activity) of the support-leg (Table 4-3, Figure 4-3).

Step-leg peak magnitude mean(SD) values ranged from: RF: 29.0(18.4) *(phase 4)* to 100.2(27.9) *(phase 2)* %; BF: 42.8(21.4) *(phase 4)* to 98.4(23.9) *(phase 3)* %; TA: 71.5(31.9) *(phase 4)* to 99.6(23.4) *(phase 2)* %; MG: 25.7(14.8) *(phase 4)* to 107.8(15.0) *(phase 1)* %. Support-leg peak magnitude mean(SD) values ranged from: RF: 49.8(18.9) *(phase 1)* to 110.2(14.4) *(phase 3)* %; BF: 23.7(18.8) *(phase 4)* to 110.1(12.9) *(phase 1)* %; TA: 35.9(19.0) *(phase 1)* to 111.0(12.3) *(phase 3)* %; MG:

24.2(29.4) *(phase 3)* to 107.0(23.4) *(phase 1)* %. Averaged time-series EMG signals are shown in Figure

4-4.

Table 4-3: Mean(SD) for the peak magnitude mean values, for each muscle and phase. Values are in %
peak of the averaged signal.

Muscle		Phase 1	Ph. 2	Ph. 3	Ph. 4	Friedman Test
R	RF	65.9(25.2) [2,4]	100.2(27.9) [1,3,4]	75.5(33.1) [2,4]	29.0(18.4) [1,2,3]	$X^2(3)=35.22$; **p<0.001**
	BF	85.9(33.5) [2,4]	70.7(25.3) [1,3,4]	98.4(23.9) [2,4]	42.8(21.4) [1,2,3]	$X^2(3)=31.86$; **p<0.001**
	TA	87.9(28.4) [2]	99.6(23.4) [1,4]	89.2(31.3) [4]	71.5(31.9) [2,3]	$X^2(3)=10.38$; **p=0.016**
	MG	107.8(15.0) [2,3,4]	74.6(14.6) [1,3,4]	85.3(26.6) [1,2,4]	25.7(14.8) [1,2,3]	$X^2(3)=49.02$; **p<0.001**
L	RF	49.8(18.9) [2,3,4]	103.9(9.2) [1,3,4]	110.2(14.4) [1,2,4]	71.0(35.6) [1,2,3]	$X^2(3)=33.47$; **p<0.001**
	BF	110.1(12.9) [2,3,4]	87.0(21.2) [1,3,4]	41.7(22.1) [1,2,4]	23.7(18.8) [1,2,3]	$X^2(3)=53.46$; **p<0.001**
	TA	35.9(19.0) [2,3]	53.2(15.9) [1,3]	111.0(12.3) [1,2,4]	46.6(39.6) [3]	$X^2(3)=40.38$; **p<0.001**
	MG	107.0(23.4) [2,3,4]	93.5(24.7) [1,3,4]	24.2(29.4) [1,2]	27.6(30.8) [1,2]	$X^2(3)=42.06$; **p<0.001**

R = right; L = left; RF = rectus femoris; BF = biceps femoris; TA = tibialis anterior; MG = medial
gastrocnemius; Ph. = phase. Superscript values refer to the phases which statistically differ from the given
phase.
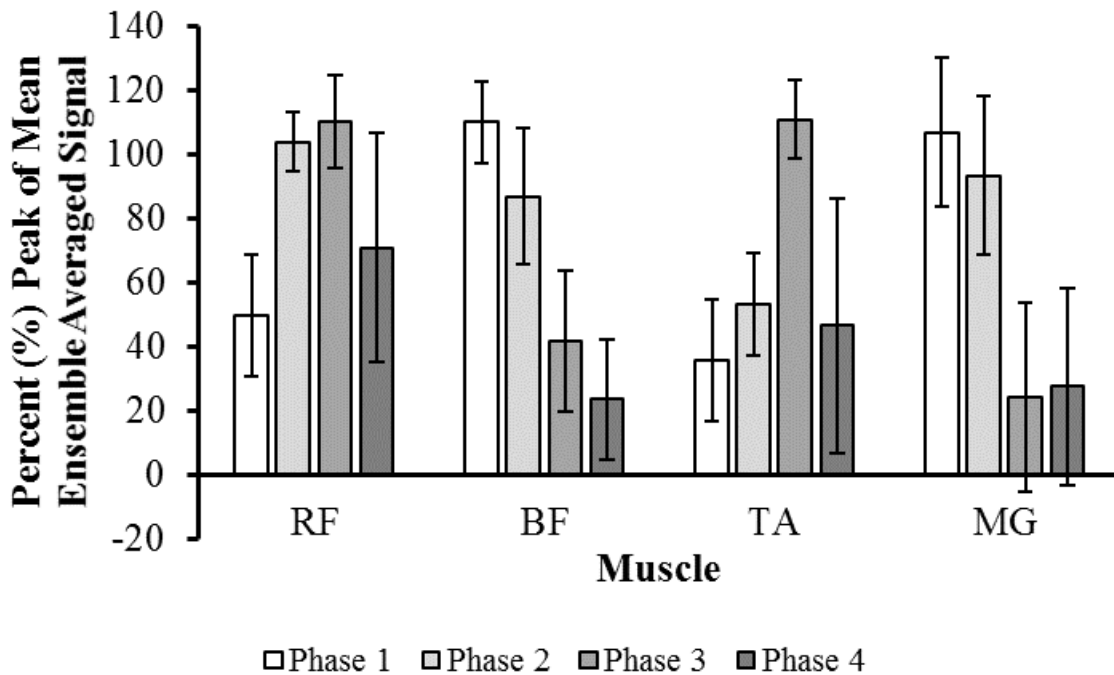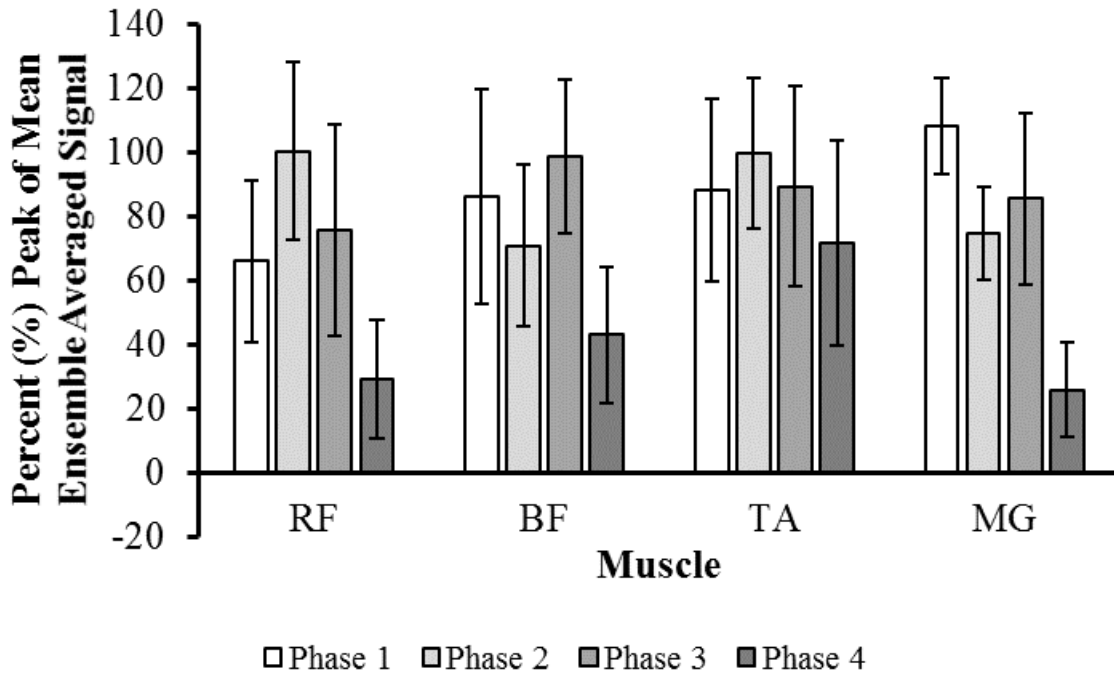
Figure 4-3: Mean(SD) for the peak magnitude mean values, for each muscle and phase. Values are in % peak of the averaged signal. The right (step-leg) is depicted in the top graph, while the left (support-leg) is depicted in the bottom graph.
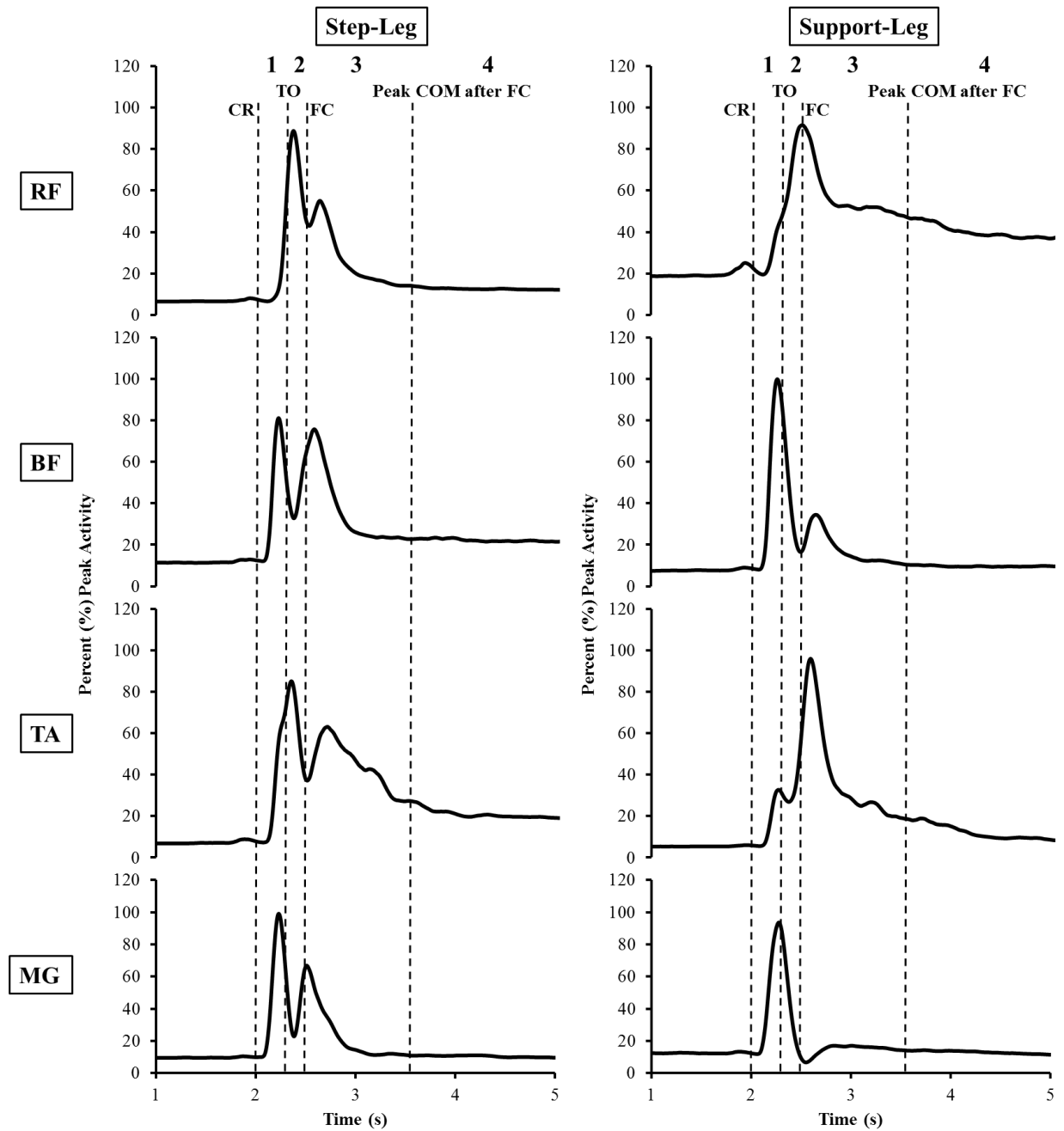
Figure 4-4: Mean time-series electromyographic signals for each muscle, of both the stepping and non-stepping legs. RF = rectus femoris; BF = biceps femoris; TA = tibialis anterior; MG = medial gastrocnemius. The phases are depicted along the top by the numbers 1, 2, 3 and 4. These curves were created by averaging the average responses of each participant.

For peak magnitude variability, main effects of phase were observed for 5 of 8 muscles (step-leg TA and MG; support-leg BF, TA, MG; $X^2(3)$=15.78–32.82; p=0.001 – p<0.001). In 3 of the 5 muscles where phase main effects were observed, phase 4 was more variable than phases 1, 2 and 3 (Table 4-4). Mean EMG traces for each participant are shown in Figure 4-5.

Table 4-4: Mean(SD) for the peak magnitude coefficient of variation values, for each muscle and phase.

Muscle		Phase 1	Ph. 2	Ph. 3	Ph. 4	Friedman Test
R	RF	22.3(4.7)	27.0(40.4)	35.8(38.4)	36.2(46.6)	$X^2(3)$=7.74; p=0.052
	BF	26.7(11.6)	36.5(48.8)	34.9(35.2)	40.7(49.1)	$X^2(3)$=2.34; p=0.505
	TA	21.2(7.0) [3,4]	30.7(48.2) [3,4]	54.9(37.9) [1,2]	56.4(31.9) [1,2]	$X^2(3)$=27.60; **p<0.001**
	MG	20.2(7.3) [4]	24.5(19.5) [3,4]	28.6(35.4) [2,4]	42.5(37.3) [1,2,3]	$X^2(3)$=20.10; **p<0.001**
L	RF	30.0(17.1)	29.5(10.9)	27.5(10.0)	29.1(16.1)	$X^2(3)$=1.13; p=0.769
	BF	19.1(7.0) [2,3,4]	22.6(7.5) [1,4]	29.9(15.1) [1,4]	39.8(19.0) [1,2,3]	$X^2(3)$=19.11; **p<0.001**
	TA	35.5(11.3) [3,4]	31.0(9.7) [3,4]	17.3(6.8) [1,2,4]	67.2(27.7) [1,2,3]	$X^2(3)$=32.82; **p<0.001**
	MG	27.3(14.9) [3,4]	28.8(15.4) [3]	41.8(19.8) [1,2]	37.2(17.4) [1]	$X^2(3)$=15.78; **p=0.001**

R = right; L = left; RF = rectus femoris; BF = biceps femoris; TA = tibialis anterior; MG = medial gastrocnemius; Ph. = phase. Superscript values refer to the phases which statistically differ from the given phase.

Lastly, the results of the peak magnitude correlations, between muscles for each phase, are presented in Table 4-5. During the landing phase, significant correlations were observed between the step-leg BF and MG (r=0.683, p≤0.01) and the support-leg BF and MG (r=0.558, p≤0.05) during phase 3. During phase 4, the step-leg RF and MG (r=0.517, p≤0.05), and the support-leg BF and MG (r=0.565, p≤0.05) were each significantly correlated.
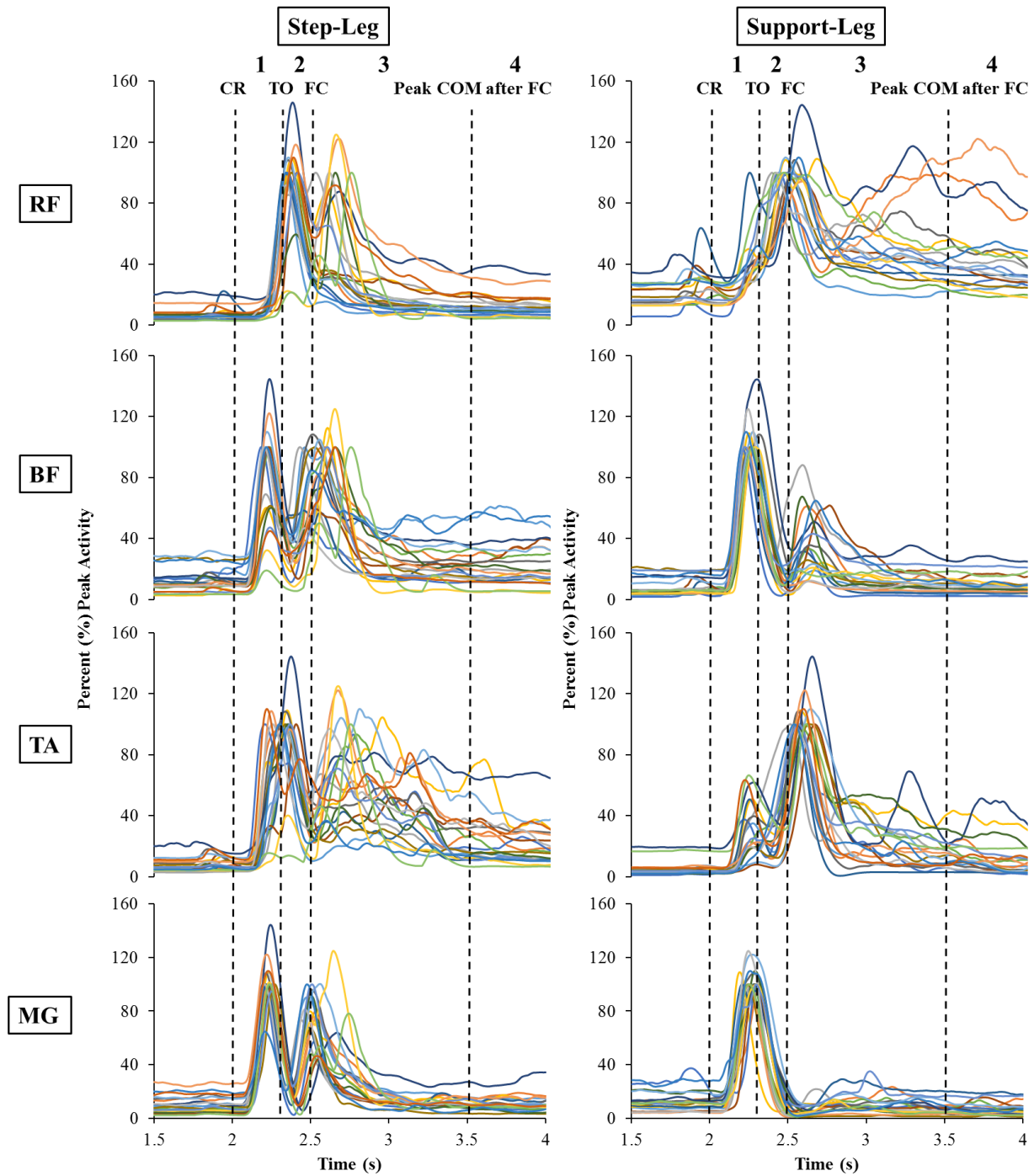
Figure 4-5: Individual participant mean time-series electromyographic signals for each muscle, of both the stepping and non-stepping legs. RF = rectus femoris; BF = biceps femoris; TA = tibialis anterior; MG = medial gastrocnemius. These curves were created by averaging each participant's data, incorporating all of the trials which were completed.

Table 4-5: Correlation coefficients for the peak magnitudes between each muscle, during each phase.

	RIGHT			LEFT		
PHASE 1	**BF**	**TA**	**MG**	**BF**	**TA**	**MG**
RF	.571*	.376	.395	.191	.304	.705**
BF	---------	.495*	.283	----------	.130	.374
TA	---------	---------	.493*	----------	----------	.198
PHASE 2	**BF**	**TA**	**MG**	**BF**	**TA**	**MG**
RF	.299	.475*	.117	-.015	.195	-.503*
BF	---------	.289	.334	----------	.433	.191
TA	---------	---------	.441	----------	----------	-.098
PHASE 3	**BF**	**TA**	**MG**	**BF**	**TA**	**MG**
RF	.012	.438	.050	-.147	.364	-.071
BF	---------	.095	.683**	----------	-.086	.558*
TA	---------	---------	.164	----------	----------	-.048
PHASE 4	**BF**	**TA**	**MG**	**BF**	**TA**	**MG**
RF	-.140	.343	.517*	.453	.451	.455
BF	---------	-.287	.009	----------	.337	.565*
TA	---------	---------	.353	----------	----------	.209

RF = rectus femoris; BF = biceps femoris; TA = tibialis anterior; MG = medial gastrocnemius; * $p \leq 0.05$; ** $p \leq 0.01$.

4.5 Discussion

The goals of this study were to assess peak lower-limb muscle recruitment patterns during reactive stepping, particularly during the landing phase, and to assess between-muscle relationships in peak magnitude within each phase. Generally, phase 4 occurred with the latest and most variable peak timing, and typically with the smallest peak magnitude. As hypothesized the peak magnitudes of the BF and MG were positively correlated during phase 3, in both the step and support-leg. Overall, these results may be helpful in understanding muscle control strategies during the landing phase, which is important for the development of appropriately targeted exercise interventions.

The primary objective of this study was to assess baseline lower-limb muscle recruitment patterns (peak timing and magnitude) over the entirety of the reactive stepping response, including after the peak COM position following FC, which is typically the kinematic event of interest in analysis of the landing

phase (Singer, et al., 2013, 2016). It was the goal that this analysis would provide further insight into the landing phase, and possible differences before and after the peak COM position after FC. Overall, the latest peak timing occurred during phase 4. During phase 3, which occurred from FC to the peak COM after FC, mean values in Table 4-1 revealed that peak timing was second earliest during this phase for each step-leg muscle, except for the MG, where peak timing during phase 3 was the earliest. These values reveal that rapid muscle activity during the initial part of the landing phase is important. Such rapidly occurring activity may be particularly important in the MG, due to the role of the plantar-flexors in generating extension moments (Pijnappels, et al., 2004, 2005; Pijnappels, et al., 2008), to resist forward rotation. Regarding the peak timing variability, for each muscle, phase 1 was significantly less variable than all other phases, except for phase 2 of the support-leg. When looking at the landing phase, phase 4 was significantly more variable than phase 3 in 63% (5 of 8) of the muscles (Table 4-2). This accompanied by the rapid peak timing values observed for phase 3 (compared to phase 4), suggests that the landing phase has two separate phases, where during phase 3, peak EMG timing is rapid and less variable compared to phase 4.

The peak magnitude for each muscle was influenced by a phase main effect. In phase 1, the step-leg MG was at its highest peak magnitude. The step-leg BF and TA also exhibited relatively large magnitudes during phase 1. In the support-leg, the MG during phase 1 exhibited its largest peak magnitude, similar to the step-leg. Additionally, the support-leg BF exhibited its largest peak magnitude during phase 1, which was likely in an hip extension role, to resist the rotational effects of the perturbation. The results regarding the BF and MG are in agreement with previous research showing the importance of these muscles in resisting forward rotation and generating a horizontal push-off (or braking) force (Pijnappels, et al., 2004, 2005; Pijnappels, et al., 2008). Overall, these early responses, in the MG specifically, are consistent with early automatic postural responses (e.g., the generation of ankle moments that typically precede reactive stepping) noted previously in response to external perturbations (McIlroy & Maki, 1993a, 1993b, 1995; Weerdesteyn, et al., 2008).

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During phase 2, the step-leg RF and TA were at their most active. The RF would be important for hip flexion to lift the step-leg, while the TA would be necessary for concentric dorsi-flexion to provide adequate toe clearance during stepping. Overall, we see the importance of the flexors (i.e., RF and TA) in the step-leg during phase 2 (Table 4-3). Regarding the support-leg, the RF was at its second highest peak magnitude during phase 2, where it helps to maintain hip height above the ground via the generation of a knee extensor moment, preventing the knee from buckling. Previously, hip height has been shown to be a modifiable feature of the reactive stepping response (Bhatt et al., 2012). Support-leg RF recruitment was accompanied by support-leg BF recruitment, likely in a hip extension role, to keep the trunk upright during the swing phase and resist the rotational effects of the perturbation. This is in agreement with early research on muscle activity during gait which suggested that the more proximal muscles of the knee and hip not only prevent collapse of the lower limb, but are also primarily responsible for correcting posture and balance of the head, arms and trunk, which together comprise the dominant mass of the body (Winter & Yack, 1987). The support-leg MG was also recruited during phase 2. During the swing phase (phase 2), the COM would be located anterior to the support-leg, which would cause a dorsi-flexion moment about the support-leg ankle. Activation of the support-leg MG would counteract this ankle dorsi-flexion by generating a plantar flexion moment, helping to keep the support-leg ankle from collapsing.

During phase 3 (the first phase of the landing phase), the step-leg BF was at its most active, while the step-leg MG was at its second most active. The BF, in its role as a hip extensor, may be acting eccentrically to help counteract the gravitational moment about the trunk. Without activation from the BF, the gravitational moment would cause forward and downward rotation of the trunk, due to the large inertia of this body segment. Previous research suggests that low-strength older adults are at a greater risk of falling during after-step falls (Pavol, et al., 2001, 2002). After-step falls have been suggested to be related to lower-extremity weakness, primarily of the hip and knee extensors, as primary factors in such falls were excessive lumbar flexion as well as buckling of the recovery limb (Pavol, et al., 2001, 2002). The large relative activation of the step-leg BF during phase 3, is in-line with the notion of the hip extensors being important after FC. The step-leg MG would also be important during phase 3 to help

move the center of pressure anteriorly, via the generation of a plantar flexion moment, which would help to arrest the COM movement. Both the hip extensors and plantar flexors have been examined previously, in the support-leg, for their role in resisting forward-rotation after a trip (Pijnappels, et al., 2004, 2005; Pijnappels, et al., 2008). Contributions from the step-leg RF and TA during phase 3 likely served to extend the knee (moving the body upwards, away from the ground), and dorsi-flex the ankle (eccentrically) to prevent the step-foot from slapping against the force-plate, respectively. The step-leg TA could also serve to bring the heel back to the floor if needed during phase 3.

In the support-leg, the peak magnitude of the RF and TA during phase 3 were larger than all of the other phases, within each muscle. The support-leg RF likely served to keep the knee-extended during landing. This recruitment could be eccentric in nature, as the gravitational effects from the perturbation would act to collapse the knee into a 'flexed' position. For the support-leg TA, its magnitude during phase 3 may be linked with that of the support-leg BF and MG during phase 1, where both of these muscles exhibited their largest peak magnitude (Table 4-3). Previous research has shown that the support-leg is important for restraining forward rotation and generating a braking force, via rapid muscles responses in the triceps surae and hamstring muscles. These muscles generate a large ankle plantar flexion moment, knee flexion moment and hip extension moment (Pijnappels, et al., 2004, 2005; Pijnappels, et al., 2008). Therefore, the use of the support-leg TA from FC onward (phase 3) will likely be closely linked with the 'effectiveness' of the support limb plantar flexors, knee flexors and hip extensors prior to FC. For instance, if the BF and MG do not adequately resist the body's forward rotation upon perturbation onset, one may continue to move anteriorly onto their support-leg toes in preparation for a second step, in which case the TA would be needed to pull the heel-back to the ground, via the generation of a dorsi-flexion moment (if using one-step for recovery). This action would help move the COM posteriorly, away from the anterior base-of-support limits, and safely between the two legs in a tandem configuration. During phase 3, the support-leg MG peak magnitude was low (relative to phases 1 and 2) as plantar-flexion from the support-leg, during the tandem stance configuration after FC, would cause the COM to move anteriorly onto the support-leg toes, which would be counter-productive to stability. Such an action would

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be useful for a second step (to propel the leg into swing), but not when using a single-step for recovery (this idea could be explored in future research). Interestingly, research of rapid gait termination in young and older adults reported that the main qualitative features of the support-leg (or stance limb) data were the reciprocal strong tibialis anterior activation and marked soleus inhibition to reduce forward momentum (Tirosh & Sparrow, 2005).

During phase 4, the step-leg BF likely continues to act in a hip extension role (possibly concentrically at this point), to help keep the trunk upright. Linking kinematic data with EMG responses in future research may provide more specific insight into the role of BF activation (e.g., concentric or eccentric) during the landing phase. Additionally, the support-leg RF peak magnitude was still relatively large during phase 4, as the knee must remain extended. However, as a whole, peak magnitudes during phase 4 were generally small when compared to the earlier three phases. This is highlighted by Figure 4-4, where each muscle appears to be at its lowest activation during this later aspect of the landing phase.

To interpret peak magnitude variability, with respect to the notion of a stereotypical response (i.e., high magnitude, low variability), the phase where each muscle exhibited its highest peak magnitude will be focused on. Accordingly, the support-leg TA during phase 3 had the lowest peak magnitude CV of all muscle-phase combinations, where this value was also significantly smaller than phases 1, 2 and 4 for the support-leg TA. This coupled with the large peak magnitude observed in this muscle suggests a stereotypical response, as dorsi-flexion to bring the support-leg heel to the floor and move the COM posteriorly would be important in single-step reactions, to arrest anterior pushing of the COM from the support-leg. Additionally, variability of the BF of the support-leg in phase 1 was lower than phases 2, 3 and 4 (and the second lowest mean CV overall) (Table 4-4). This coupled with the high magnitude of the support-leg BF observed in phase 1 suggests that hip extension in the support-leg following cable-release (CR) occurs with a relatively consistent magnitude to resist the rotational effects of the perturbation, particularly about the trunk which is the largest segment of the body. Overall, peak magnitudes during phase 3 were always larger than that of phase 4 (except for the MG of the support-leg) (Table 4-3). Additionally, for the step-leg, peak magnitudes during phase 4 were significantly smaller than every other

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phase, except for phase 1 of the TA. Along with the peak timing results, this adds to the notion that the landing phase is comprised of two phases: one phase immediately following FC, which is dynamic, requiring rapid muscle recruitment with large relative peak magnitudes, and a second quasi-static phase (following the peak COM position after FC), where peak timing is slower, more variable, and the peak magnitudes are smaller.

While the values presented in Figures 4-3 and 4-4 provide interesting insights into mean muscle activation during the reactive stepping response, the subject-specific mean\averaged responses presented in Figure 4-5 may provide information regarding which of the muscle responses are 'stereotypical' and which are variable between participants. The MG of the both the step and support-leg appear to be very similar across participants in terms of the number of peaks, and timing (specifically the first peak of the step-leg). Interestingly, the support-leg BF was also similar across participants. Together, the common responses observed in the support-leg BF and MG across participants could suggest an important role for these muscles during the reactive stepping response, a notion which has been reported previously (Pijnappels, et al., 2004, 2005; Pijnappels, et al., 2008). Alternatively, perhaps the more variable muscles are important for balance control during the reactive stepping response. When one examines Figure 4-5, it appears as though the muscles of the anterior chain (RF and TA) were highly variable. Specifically, the step-leg TA and support-leg RF showed large peak magnitude variability between-subjects during the 3 to 4 second period of Figure 4-5, which could suggest that these muscles are particularly important for control during the landing phase. Due to the between-subject variability in lower-limb muscle activation, future studies should explore the link between landing phase kinematics\kinetics and muscle activation.

Regarding the correlations between the peak magnitudes, the landing phase (phases 3 and 4) will be focused on during the Discussion. In agreement with the hypothesis, during phase 3 the BF and MG of the step-leg positively correlated with each other. The step-leg BF and MG, similar to what has been observed for the support-leg, are likely very important for resisting forward rotation via large ankle plantar flexion, knee flexion and hip extension moments (Pijnappels, et al., 2004, 2005; Pijnappels, et al., 2008). As such, clinicians may wish to focus on these two muscles. Similar correlations were also

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observed in the support-leg. However, the correlation was not as strong as observed in the step-leg ($r =$ 0.683 vs 0.558) (Table 4-5), which may be because at FC, the support-leg configuration was typically extended at the knee, which reduces the moment arm of the BF for hip extension. During phase 4, the step-leg RF and MG were positively correlated (as hypothesized), however the step-leg BF and TA were not. Yet, in the support-leg, the BF and MG were again positively correlated. Contractions by the step-leg RF and MG would serve to move the body backward towards the support limb (Table 4-5), while (as discussed) the support-leg BF and MG provide hip extension and plantar flexion moments required to keep the body away from the anterior limits of stability. However, if after a single reactive step, the support-leg is configured such that the participant is on their toes, continued plantar-flexion could be destabilizing. Overall, the correlation results could aid researchers in training recruitment patterns in older adults, via biofeedback programs etc.

As with Study 1, the results of this study must be interpreted with caution, due to the predictable nature of the tether-release perturbation direction. Participants were also instructed to respond with a single forward step, which may have promoted response pre-planning. Study 2 was also limited by the fact that maximal voluntary contractions were not collected. The reason for this was because we were most interested in differences across phases of the reactive stepping response, and not necessarily between muscles. Further, due to the challenging nature of the tether-release task, we did not want to fatigue the participants unnecessarily prior to the reactive stepping trials, as the participants also completed the active range-of-motion and strength tasks described in Study 1, section 3.3.1. Therefore, we cannot compare EMG magnitude between muscles, nor could measures such as a co-contraction index be calculated to gain insight into agonist and antagonistic recruitment. Nonetheless, the peak magnitudes were focused on to gain insight into whether the relative contribution of each muscle changed from phase-to-phase. Second, no muscles which are related to medio-lateral control, such as the gluteus medius, were collected due to constraints caused by the safety harness and EMG belt which all participants had to wear. As the tether-release perturbation primarily occurs in the anterior-posterior direction, collecting EMG activity from primarily anterior-posterior muscles was logical. Further, although the objective of the study

was to quantify baseline EMG during the landing phase of reactive stepping, the fact that only young adults were collected means that no direct insight or conclusions can be made regarding lower-limb EMG in older adults, during the landing phase. Finally, the decision to separate each trial, for each participant, into four phases (or bins) was also associated with limitations. Although the four phases were used to determine both peak timing and peak magnitude, muscle activation onset (after CR) was not calculated. Therefore, the timing of the earliest leg muscle activation was not captured in the current study. Additionally, it is possible that a 'peak' in a given trial could have spanned multiple phases. As such, it is possible that some peak magnitudes may actually represent the highest value during the ascending or descending aspect of a larger 'true' peak, achieved in a separate phase (e.g., Figure 4-4 – MG of the step-leg from TO to FC). Lastly, Figure 4-4 represents averaged data for each muscle, which includes data from all 20 participants. This makes it unclear whether the patterns shown represent 'real' common responses, or if the patterns are a result of differences in event timing (TO, FC, peak COM after FC), between-participants, and potentially within-trials for each participant. In an attempt to show participant specific data, Figure 4-5 was also presented, with participant specific averaged time-series data for each muscle. Nonetheless, these figures were not meant to be a detailed analysis, but instead to show trends which generally agree with the data which was used for the statistical analyses, calculated on a trial-by-trial basis for each participant (Tables 4-1 to 4-4). Despite the limitations of the current study, it was the first to quantify EMG recruitment patterns in the lower-limbs during the landing phase of reactive stepping.

In conclusion, this study provided a basic science characterization of the leg muscle recruitment patterns in young adults during reactive stepping. Using this information, researchers can next assess how recruitment patterns change with age or specific pathology, while clinicians can use this baseline young adult data to target phases of reactive stepping by strengthening the muscles which were most active during the landing phase, such as the step-leg BF (which was correlated with the step-leg MG peak magnitude) and the support-leg RF and TA. Moving forward (Chapter 5), research is needed to quantify

how arm movement and wide stepping influence balance control during the landing phase, as these

movements are often used by older adults.

5. STUDY THREE – INFLUENCE OF WIDE STEPPING AND ARM MOVEMENT ON BALANCE CONTROL DURING REACTIVE STEPPING: A FOCUS ON STABILITY AFTER FOOT-CONTACT

5.1 Chapter Overview

Numerous studies have shown older adults to use wide steps and large arm movement during balance recovery. However, it is unclear how these strategies influence reactive stepping, particularly after foot-contact (FC). Therefore, the purpose of this study was to examine the influence of wide stepping and restricted arm movement on landing phase control. Twenty young adults and 16 older adults participated in three conditions: 1) preferred stepping; 2) wide stepping, and; 3) restricted arm movement. Full body kinematics were used to quantify the peak center of mass (COM) displacement after FC, in the anterior-posterior (AP) and medio-lateral (ML) directions. Wide-stepping resulted in a larger AP COM displacement compared to both other conditions ($p \leq 0.001$). Older adults also exhibited a larger peak AP COM displacement after FC ($p=0.003$). In the ML direction, older adults had a greater ML displacement during the preferred stepping condition ($p=0.018$) and tended to have a greater displacement in the restricted arm movement condition ($p=0.062$). Medio-lateral COM displacement was largest during wide stepping and second largest in the restricted arm movement condition ($p \leq 0.001$). In the AP direction, the restricted arm movement condition was significantly less variable than both other conditions ($p=0.010$-0.050), while wide stepping was most variable in the ML direction ($p<0.001$). Wide stepping resulted in the largest AP and ML body movement after FC, where the former could be counterproductive to increased stability. Second, restricted arm movement resulted in larger ML body movement after FC, compared to preferred stepping, suggesting that arm movement is important for ML control.

5.2 Introduction

Despite recent research revealing deficits in older adults (Singer, et al., 2016), it is unclear how specific movements/compensatory strategies influence stability during the landing phase. During forward and backward reactive stepping, older adults often exhibit a laterally directed step placement (McIlroy &

Maki, 1996; Rogers, et al., 2001; Schulz, et al., 2005; Troy, et al., 2008). Further, during normal or narrow walking, older adults use wider steps compared to young adults (Dean, et al., 2007; Schrager, et al., 2008), with greater stride width variability (Dean, et al., 2007; Grabiner, et al., 2001). An increased stride width in older adults has also been reported during rapid gait termination (Menant, et al., 2009). During volitional lateral stepping, older adults' medio-lateral center of pressure velocity remains above baseline levels 30-seconds post-step (Porter & Nantel, 2015), suggesting age-related restabilisation deficits during lateral stepping.

Older adults are also more reliant on the use of their arms for assistance during balance recovery after a postural perturbation (Maki, et al., 2000), despite being unable to initiate arm movements as rapidly (Allum, et al., 2002; Maki, et al., 2001; Mansfield & Maki, 2009; Weaver, et al., 2012), or to the same extent as the young (Allum, et al., 2002). Without restricting arm movement during the reactive stepping task, Singer (2012) and Singer et al. (2016) observed deficits during the landing phase in older adults, compared to young adults. Although not quantified, the older adults exhibited larger arm abduction after step initiation (Singer, 2012). Logically, if older adults are more reliant on their arms, restraining the arms should amplify previously reported age-related differences in landing phase balance control.

The objective of this study was to determine how wide stepping and restricted arm movement influence balance control after foot-contact (FC) during forward reactive stepping, in young and older adults. Both mean values and standard deviations (SD) were compared, as existing research has shown increased variability in older adults during the landing phase (Singer, et al., 2013, 2016). It was hypothesized that: 1) wide stepping would result in the largest and most variable peak anterior-posterior (AP) and medio-lateral (ML) center of mass (COM) displacement after FC; 2) restricted arm movement would result in a larger and more variable peak AP and ML COM displacement after FC, compared to the preferred stepping trials; and 3) compared to the young adults, older adults would exhibit a larger and more variable peak AP and ML COM displacement after FC during all experimental conditions. However, the between-group differences would be largest during the wide stepping and restricted arm

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movement conditions. Understanding how wide stepping and arm movement influence landing phase balance control will aid researchers and clinicians who wish to improve reactive stepping performance in older adults.

5.3 Methods

5.3.1 Experimental Protocol

Twenty young adults (age: 22.4(3.1) y; height: 1.7(0.1) m; mass: 74.0 (20.6) kg; 10 females) and 16 older adults (age: 69.6(4.0) y; height: 1.6(0.1) m; mass: 67.6(12.4) kg; 13 females) participated in Study 3. These participants were part of a larger data collection, described in Chapter 3, section 3.3.1. Ethics clearance was obtained from the University of Waterloo Human Research Ethics Committee prior to study commencement. Telephone interviews were conducted to ensure participant eligibility (Appendix 1). All participants provided their informed consent prior to completing any aspect of the study.

Twelve cameras (Optotrak Certus, Northern Digital Incorporated, Waterloo, Ontario, Canada) were used to collect kinematic data at 64Hz, while three force-plates (BP 5050 (x2), Bertec, Columbus, Ohio, USA, and OR6-7 (x1), Advanced Mechanical Technology Inc., (AMTI) Watertown, MA, USA) were used to collect ground reaction forces and moments at 2048 Hz. The two Bertec force-plates were arranged side-by-side (Figure 5-1 and Figure 5-2). The AMTI force-plate was where the participants stepped onto, using their right leg. The tether supporting each participant's body weight was located in-line with a load cell (MLP-300-CO, Transducer Techniques, Temecula, CA) which was sampled at 2048 Hz and rated for up to 136 kg. The tether was connected to a metal frame via an electromagnet (AEC Magnetics, Cincinnati, OH, USA).
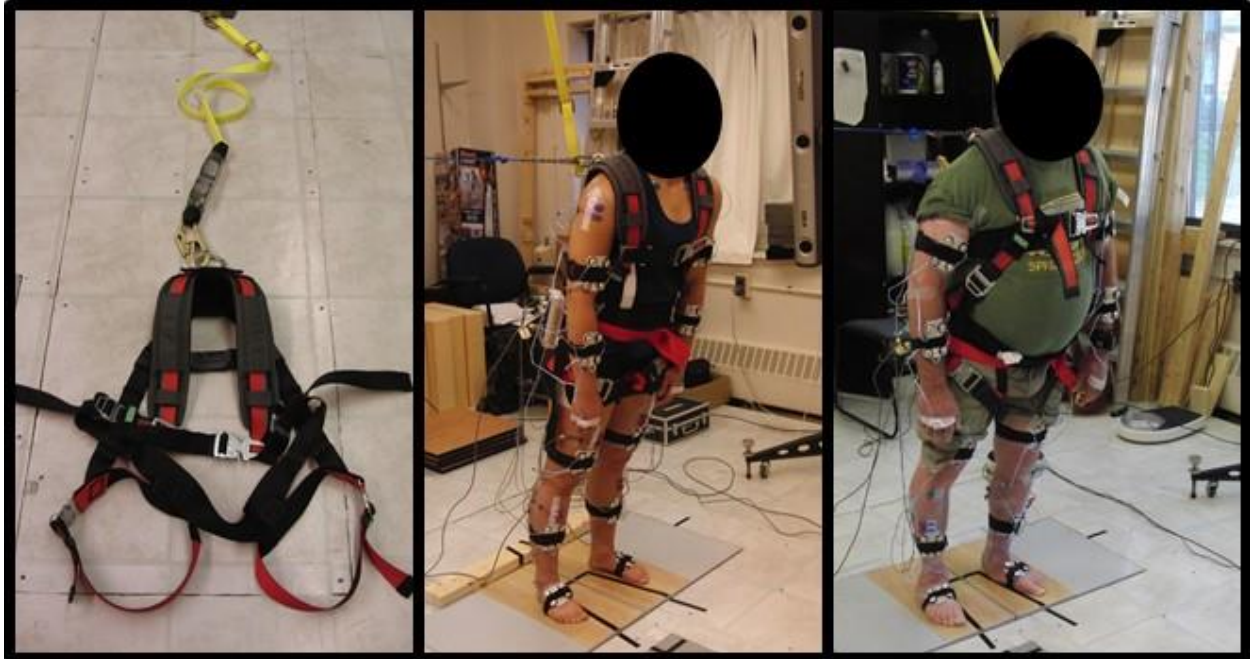
Figure 5-1: On the left is the safety harness which participants wore, along with the safety-tether which was connected to the ceiling to prevent participants from falling to the ground. In the middle, a young adult participant, and on the right, an older adult participant are depicted wearing the harness, along with the safety tether. The two Bertec force-plates are also depicted in both photos.
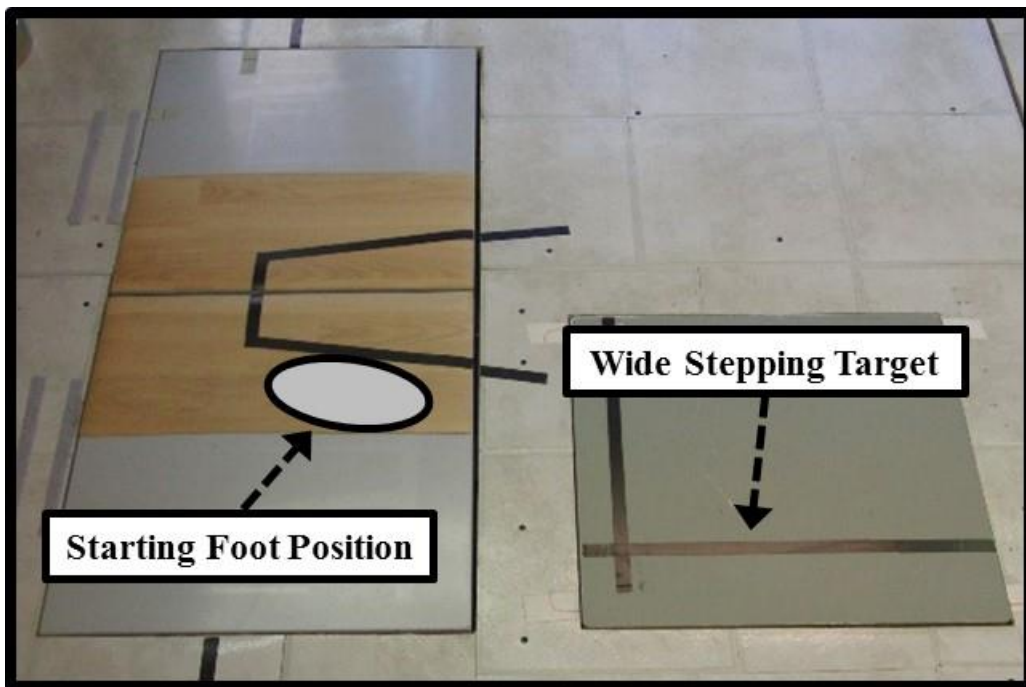
Figure 5-2: The layout of the three force-plates is depicted. The wide stepping target is also displayed (black electrical tape), along with the participant's initial foot position.

In the current study, ankle muscle electromyography and full body kinematics were collected. Electromyographic data was sampled at 2048 Hz using a differential amplifier, with a hardware band-pass filter of 10-1000 Hz, a common mode rejection ratio of 115dB at 60 Hz (Bortec Biomedical, Calgary, AB) and disposable, self-adhesive Ag/Ag-Cl electrodes which were placed bilaterally on the tibialis anterior (TA) and medial gastrocnemius (MG). All motion capture data and analog-to-digital converted signals were synchronized using First Principles software (Northern Digital Incorporated, Waterloo, Ontario, Canada). The analog-to-digital converter included a 16-bit card. Kinematics were measured using a whole-body marker set. Rigid clusters of four markers were placed on the locations depicted in Appendix 2. Additionally, using a digitizing probe (Northern Digital Incorporated, Waterloo, Ontario, Canada) "imaginary" markers were digitized bilaterally at anatomically relevant locations (Appendix 2). The laboratory global coordinate system was defined in accordance with ISB recommendations (Wu & Cavanagh, 1995).

Briefly, this data collection was part of a larger experiment, which has previously been described in Chapter 3 (section 3.3.1) (Figure 5-1). The quiet standing trials, the tether-release method used to provide the postural perturbations, and all of the pre-release conditions that participants were asked to meet have been detailed in Chapter 3 (section 3.3.1) and Appendix 1. In addition to the tether-release trials collected as part of Chapter 3 (i.e., 5 practice trials, 10 preferred stepping trials) participants also completed two experimental conditions involving reactive stepping after tether-release perturbations: 1) wide stepping, and 2) restricted arm movement. During the wide stepping trials, participants were instructed to respond to each trial with a single-step of their right leg, with a width that was aligned with a black piece of electrical tape located near the lateral border of the target stepping force-plate (Figure 5-2). From the standardized start position, the wide stepping target equated to a step width of 0.31 m. However, if the participant's foot was touching the tape, this was accepted as a 'successful' trial. This distance was the maximum available distance, based on the force-plate participants stepped onto. These trials were repeated until five 'successful' trials were completed. No additional restrictions were placed on

participant step length (aside from the AMTI force-plate). Participants started each trial with their arms at

their sides, but no restrictions were placed on arm movement after tether-release.

For the restricted arm movement trials, participants were instructed to respond to each tether-

release perturbation with a single (right leg) step, and to not use their arms. Participants were instructed to

start each trial by holding onto their shorts, and to keep holding their shorts throughout the recovery

process. Trials where a participant was unable to adhere to the instruction were repeated. No additional

restrictions were placed on participant step length or step width.

Regardless of the experimental condition, participants were instructed to maintain their final

position for approximately 10 seconds once they regained their stability, to allow for the landing phase to

be captured (Singer, 2012). The practice trials, and preferred stepping condition were always completed

first (described in Chapter 3, section 3.3.1), after which the wide stepping and restricted arm movement

conditions were block randomized across participants.

5.3.2 *Data Analysis*

All kinematic data was low-pass filtered using a 2nd order, dual-pass, Butterworth filter with a

cut-off frequency of 6 Hz (Graham, et al., 2015; Singer, et al., 2016). An estimate of the whole body

center of mass (COM) was calculated using the filtered kinematic data and the anthropometric tables of

de Leva (1996) for the young adult participants and Dempster (1955) (as displayed in Winter (2009)) for

the older adults. To determine the hip and shoulder joint centers, the methods of Weinhandl and

O'Connor (2010) and Nussbaum and Zhang (2000), respectively, were used. Next, the position of the

COM, in the AP and ML directions, was calculated at the following time points: 1) toe-off, 2) FC and, 3)

the peak COM position after FC. In each instance, to calculate COM displacement, the COM was

referenced to the mean starting COM value, which was calculated from the start of the trial (frame 1) to

one frame before cable-release. The direction of the peak ML COM position after FC was determined

based on the stepping leg, which was always the right leg (i.e., peak right (+Z) position). Anterior-

posterior and ML stability margins were calculated by subtracting the COM position from the tip of the

right big toe and the right 5th metatarsal, in the AP and ML directions, respectively. Stability margins were calculated at the point of the maximum AP and ML COM positions after FC.

Force-plate data was also low-pass filtered using a 2nd order, dual-pass Butterworth filter with a cut-off frequency of 50 Hz (Singer, et al., 2016). From the force-plates, the following temporal events were defined: toe-off of the right (stepping) leg and FC of the right leg. Toe-off was defined as the time point when the vertical force under the right leg fell below 10 N (Sparrow & Tirosh, 2003), while FC was defined as the time point when the vertical force signal of the force-plate (which participants stepped onto) exceeded, and remained above 10 N (Sparrow & Tirosh, 2003). Cable-release was calculated using the data from the load cell located in-series with the tether. This data was low-pass filtered using a 2nd order, dual-pass Butterworth filter, with a cut-off frequency of 3 Hz (Wright, et al., 2014). The point at which cable-release occurred was defined in accordance with previous research (Graham, et al., 2015). Specifically, cable-release was defined as a 20% reduction in force in the load cell located in-series with the tether.

Step length and width were calculated using the COM of the right foot. The difference in the position of the right foot COM between FC and cable-release was calculated as the step length (AP) and step width (ML), respectively. Both of these values were also normalized by dividing by each participant's leg length (Graham, et al., 2015). Stability margins and step length/width were calculated as secondary variables to aid in the interpretation of the COM displacement mean and SD values.

To compare participant position prior to cable-release, multiple variables were calculated. Mean tether-load (during the lean), mean vertical force under the right and left feet and the mean start AP and ML COM positions were all calculated from the start of the trial, to one frame prior to cable-release. Similarly, mean AP and ML stability margins were also calculated from the start of the trial to one frame prior to cable-release. All stability margins were calculated with respect to the right foot.

5.3.3   *Statistical Analyses*

To examine the effects of movement condition and age-group on the size and variability of the peak COM displacement after FC, stability margins and step length and width, 3 x 2 mixed model

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ANOVAs were conducted. Regarding the peak COM displacement after FC, the ANOVAs were conducted using the mean values, as well as the SD values, which represented within-subject trial-to-trial variability, computed for each subject, in each group. Secondary analyses evaluated the initial lean-position metrics with the same mixed-model ANOVA approach. For each ANOVA, the within-subject factor consisted of movement condition (preferred stepping vs. wide stepping vs. restricted arm movement), while the between-subject factor consisted of age-group (young vs. older adults). Significant main or interaction effects were explored using post-hoc Fisher's least significant difference (LSD) paired or independent samples t-tests, as appropriate. All statistical analyses were conducted using SPSS (v.21, IBM Corporation, New York, USA). Experiment wide statistical significance was set at $p \leq 0.05$.

5.4   Results

Peak COM Displacement after FC:

In the AP direction, there was a condition main effect ($F(2,68)=28.10$; $p<0.001$), where the wide-stepping condition resulted in a larger COM displacement compared to both other conditions ($p<0.001$), regardless of age-group. A main effect of age was also observed, where the older adults exhibited a larger overall peak AP COM displacement after FC ($F(1,34)=0.34$; $p=0.003$) (Table 5-1; Figures 5-3 and 5-4). In the ML direction, a condition x age group interaction was observed ($F(2,68)=5.48$; $p=0.006$), where older adults had a greater ML displacement during the preferred stepping condition ($p=0.018$) and tended to have a greater displacement in the restricted arm movement condition ($p=0.062$), compared to the young adults (Table 5-2; Figure 5-5).

Table 5-1: Mean(SD) peak AP COM displacement after foot-contact, for each condition and age group.

Group	AP Peak COM Displacement after Foot-Contact			
	Preferred	Wide Stepping	Restricted Arms	Average (mm)
Young (mm)	227.9(35.0)	253.4(40.5)	230.4(39.8)	**237.2(39.6)** [A]
Older (mm)	263.8(34.1)	295.9(28.7)	257.6(36.2)	**272.4(36.6)** [A]
Average (mm)	**243.9(38.6)**	**272.3(41.3)**[C]	**242.5(40.1)**	----------------------

[C] $p<0.001$, compared to both the preferred and restricted arm movement conditions; [A] $p=0.003$; AP = anterior-posterior; COM = center of mass.
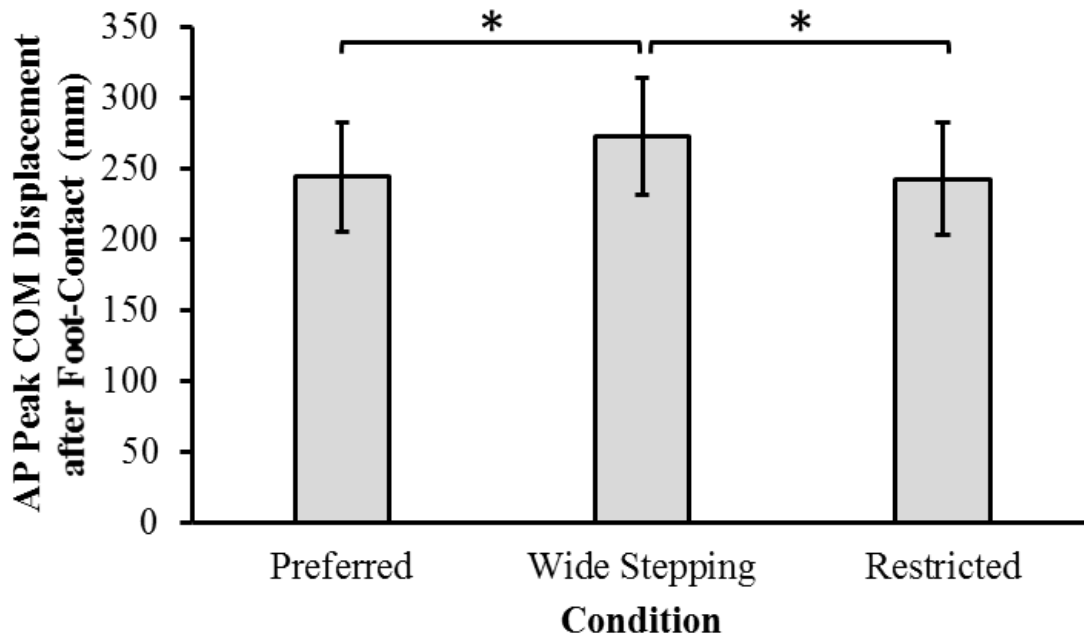
Figure 5-3: Mean(SD) values for the anterior-posterior (AP) peak center of mass (COM) displacement after foot-contact during the three movement conditions. Values have been collapsed across age groups. Restricted refers to the restricted arm movement condition. * $p\leq0.05$.
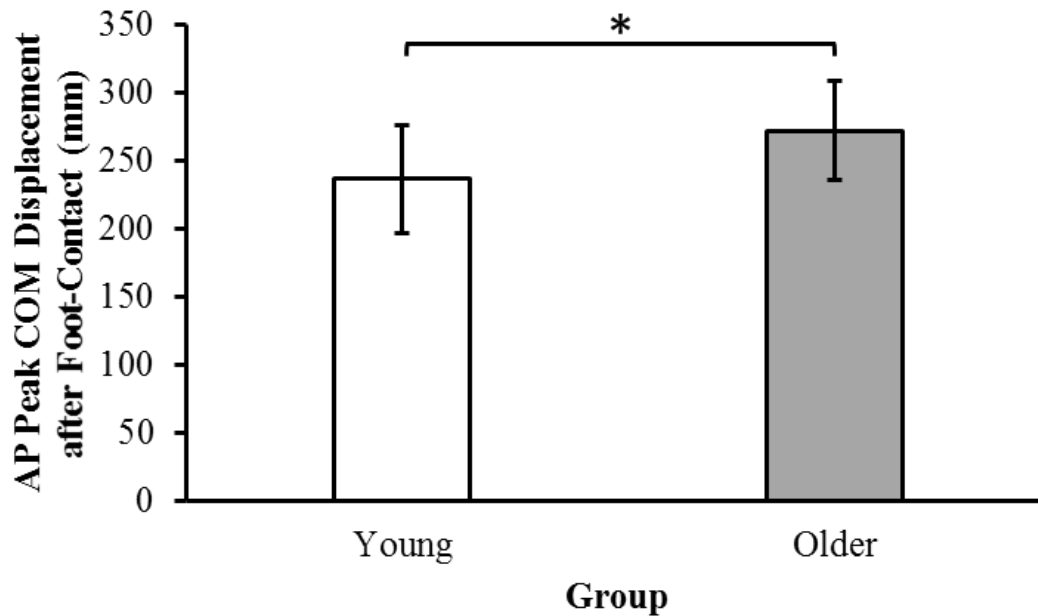
Figure 5-4: Mean(SD) values for the anterior-posterior (AP) peak center of mass (COM) displacement after foot-contact for the young and older adults. Values have been collapsed across movement conditions. * $p \leq 0.05$.

Table 5-2: Mean(SD) <u>peak ML COM displacement after foot-contact, for each condition and age group.</u>

	ML Peak COM Displacement after Foot-Contact			
Group	**Preferred**	**Wide Stepping**	**Restricted Arms**	**Average (mm)**
Young (mm)	69.8(17.7)[CA]	191.6(32.3)	82.9(23.9)[CA*]	**114.8(60.4)**
Older (mm)	87.4(24.7)[CA]	181.5 (33.8)	98.6 (24.7)[CA*]	**122.5(50.5)**
Average (mm)	**77.6(22.6)[C]**	**187.1(32.9)[C]**	**89.9(25.1)[C]**	---------------------

[C] $p \leq 0.001$; [CA] $p=0.018$; [CA*] $p=0.062$; ML = medio-lateral; COM = center of mass.
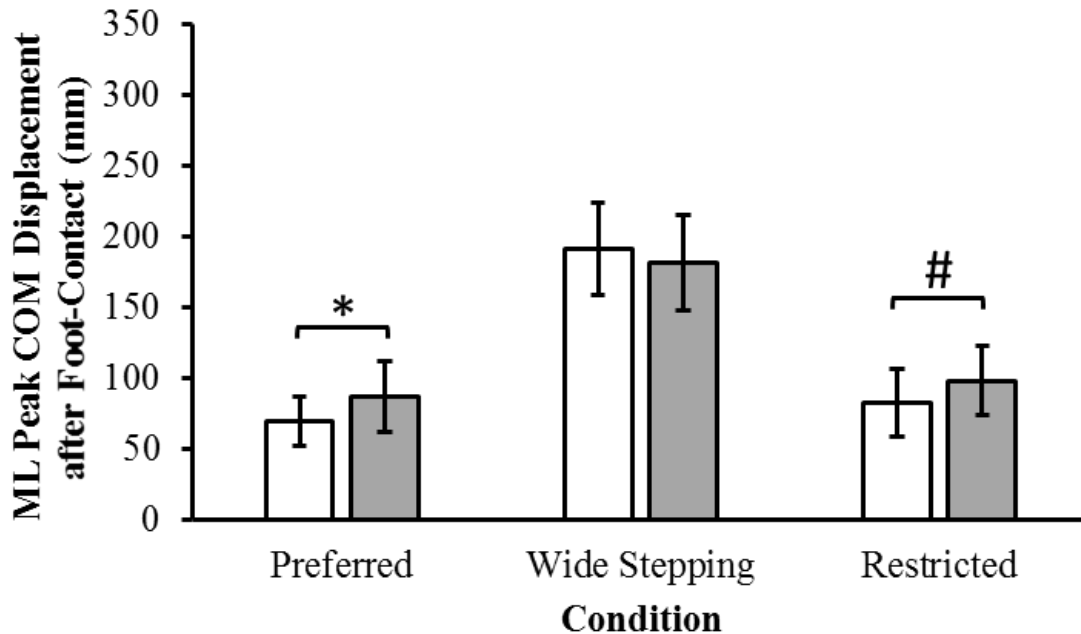
Figure 5-5: Mean(SD) values for the medio-lateral (ML) peak center of mass (COM) displacement after foot-contact. Restricted refers to the restricted arm movement condition. * $p \le 0.05$; # $0.05 < p \le 0.10$.

Variability (SD) in the AP direction was influenced by a condition main effect ($F(2,62)=4.94$; $p=0.010$), where the restricted arm movement condition was significantly less variable than both the preferred stepping ($p=0.050$) and wide stepping conditions ($p=0.010$). In the ML direction, the SD was also influenced by a condition main effect ($F(2,62)=14.08$; $p<0.001$), where both the preferred stepping ($p<0.001$) and the restricted arm movement conditions ($p<0.001$) were less variable than the wide stepping condition (Table 5-3).

Table 5-3: Mean(SD) values reflecting trial-to-trial variability for each condition, in each direction.

Dependent Variable	Preferred	Wide Stepping	Restricted Arms
AP Peak COM Disp. after FC (mm)	19.1(8.2)	21.7(9.8)	14.8(9.1)^C
ML Peak COM Disp. after FC (mm)	11.8(3.8)	23.5(15.7)^C	10.7(8.7)

$^C p \le 0.05$, compared to all other conditions in the given movement direction.

Stability Margin (at the Peak COM Position after FC):

The AP stability margin was influenced by a condition main effect ($F(2,68)$=13.96; $p<0.001$), where wide stepping resulted in a larger stability margin than both the preferred ($p<0.001$) and restricted arm movement conditions ($p<0.001$). A trend of age was also observed ($F(1,34)$=2.89; p=0.099), where older adults tended to exhibit a smaller stability margin vs. the young adults (Table 5-4). In the ML direction the stability margin was influenced by a condition x age group interaction ($F(2,68)$=7.88; p=0.001). Post-hoc analysis revealed that during the wide stepping condition, the older adults had a smaller ML stability margin at the peak COM position, compared to the young adults (p=0.017) (Table 5-5).

Table 5-4: Mean(SD) AP stability margin at the peak COM after FC, for each condition and age group.

	AP Stability Margin at Peak AP COM Position after Foot-Contact			
Group	**Preferred**	**Wide Stepping**	**Restricted Arms**	**Average (mm)**
Young (mm)	306.3(34.6)	337.5(38.8)	316.5(35.0)	**320.1(37.9)[A]**
Older (mm)	300.7(30.6)	311.6(27.3)	297.3(28.8)	**303.2(29.0)[A]**
Average (mm)	**303.8(32.5)**	**326.0(36.2)[C]**	**308.0(33.4)**	----------------------

[C] $p<0.001$, compared to both the preferred and restricted arm movement conditions; [A] p=0.099; AP = anterior-posterior; COM = center of mass; FC = foot-contact.

Table 5-5: Mean(SD) ML stability margin at the peak COM after FC, for each condition and age group.

	ML Stability Margin at Peak AP COM Position after Foot-Contact			
Group	**Preferred**	**Wide Stepping**	**Restricted Arms**	**Average (mm)**
Young (mm)	165.2(17.7)	265.7(32.6)[CA]	183.1(24.6)	**204.7(50.9)**
Older (mm)	170.1(16.0)	241.6(22.3)[CA]	179.5(21.7)	**197.1(37.7)**
Average (mm)	**167.4(16.9)[C]**	**255.0(30.6)[C]**	**181.5(21.1)[C]**	----------------------

[C] $p\leq0.001$; [CA] p=0.017; ML = medio-lateral; COM = center of mass; FC = foot-contact.

Step Length and Width:

Step length was influenced by a condition main effect ($F(2,68)$=57.70; $p<0.001$), and an age main effect ($F(1,34)$=13.51; p=0.001) (Figures 5-6 and 5-7), where the wide stepping condition resulted in longer step lengths than both the other conditions ($p<0.001$). Additionally, older adults took longer steps

regardless of condition (576.2(49.1) mm vs. 538.8(45.9) mm. Similar results were observed for normalized step length (Table 5-6). Step width was influenced by a condition x age group interaction ($F(2,68)=8.56$; $p<0.001$) (Figure 5-8). Post-hoc analysis revealed that during the preferred stepping condition, older adults took wider steps (58.5(25.9) mm vs. 37.8(16.7) mm) ($p=0.007$), but tended to take narrower steps during the wide stepping condition (218.6(49.9) mm vs. 245.2(33.0) mm) ($p=0.063$), compared to the young adult group. Normalized step width was influenced by a condition x age group interaction ($F(2,68)=4.27$; $p=0.018$). Post-hoc analysis revealed that during the preferred stepping condition, older adults (7.8(3.5) %) took wider normalized steps than young adults (4.8(2.4) %) ($p=0.004$), and tended to take wider normalized steps during the restricted arm movement condition (11.0(4.5) % vs. 8.2(4.1) %) ($p=0.057$) (Table 5-7).

Table 5-6: Mean(SD) step length values, for each movement condition and age group.

	Step Length			
Group	**Preferred**	**Wide Stepping**	**Restricted Arms**	**Average (mm)**
Young (mm)	514.2(29.6)	582.2(42.6)	520.0(29.4)	**538.8(45.9)[A]**
Older (mm)	558.9(43.3)	614.3(35.8)	555.4(45.6)	**576.2(49.1)[A]**
Average (mm)	**534.0(42.2)**	**596.5(42.4)[C]**	**535.7(41.0)**	---------------------
	Normalized Step Length			
Group	**Preferred**	**Wide Stepping**	**Restricted Arms**	**Average (%)**
Young (%)	63.5(6.4)	71.9(7.7)	64.2(6.6)	**66.5(7.8)[A]**
Older (%)	74.4(5.9)	81.8(6.2)	74.0(7.5)	**76.7(7.4)[A]**
Average (%)	**68.3(8.2)**	**76.3(8.6)[C]**	**68.6(8.5)**	---------------------

[C] $p<0.001$, compared to both the preferred and restricted arm movement conditions; [A] $p\leq0.001$.
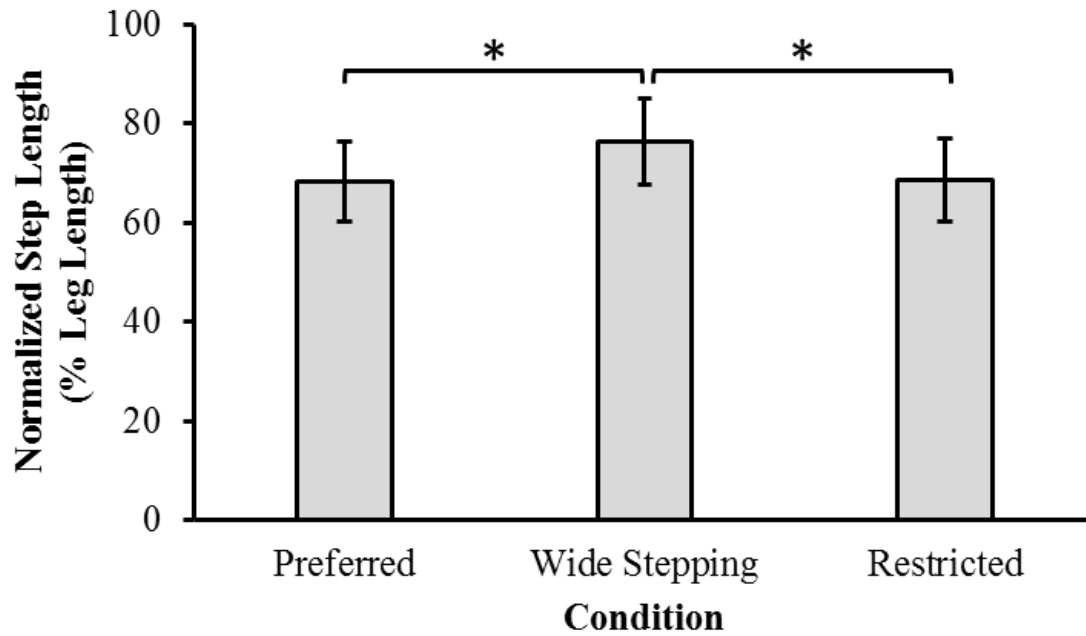
Figure 5-6: Mean(SD) values for the normalized step length during the three movement conditions. Values have been collapsed across age groups. Restricted refers to the restricted arm movement condition. * $p \leq 0.05$.
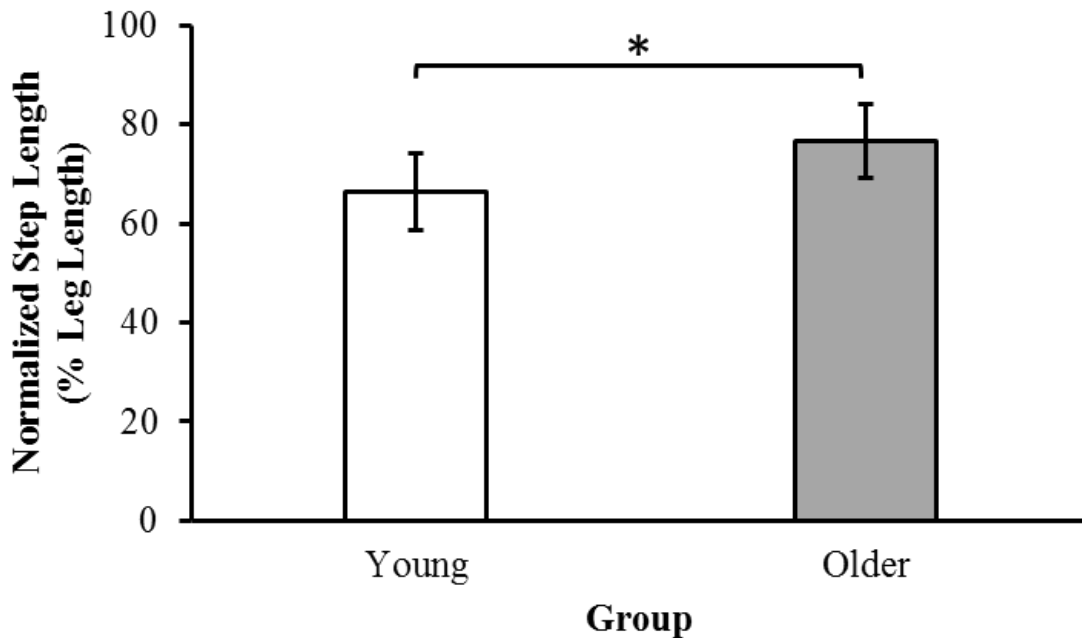
Figure 5-7: Mean(SD) values for the normalized step length for the young and older adults. Values have been collapsed across movement conditions. * $p \leq 0.05$.

Table 5-7: Mean(SD) step width values, for each movement condition and age group.

	Step Width			
Group	**Preferred**	**Wide Stepping**	**Restricted Arms**	**Average (mm)**
Young (mm)	37.8(16.7)^{CA}	245.2(33.0)^{CA*}	65.7(32.3)	**116.2(96.8)**
Older (mm)	58.5(25.9)^{CA}	218.6(49.9)^{CA*}	81.5(30.9)	**119.5(80.1)**
Average (mm)	**47.0(23.4)^{C}**	**233.4(42.9)^{C}**	**72.7(32.2)^{C}**	---------------------
	Normalized Step Width			
Group	**Preferred**	**Wide Stepping**	**Restricted Arms**	**Average (%)**
Young (%)	4.8(2.4)^{CA}	30.4(5.5)	8.2(4.1)^{CA*}	**14.5(12.2)**
Older (%)	7.8(3.5)^{CA}	29.1(6.8)	11.0(4.5)^{CA*}	**16.0(10.7)**
Average (%)	**6.1(3.3)^{C}**	**29.9(6.0)^{C}**	**9.4(4.4)^{C}**	---------------------

^{C} $p \leq 0.001$; ^{CA} $p=0.004 - 0.007$; ^{CA*} $p=0.057 - 0.063$.
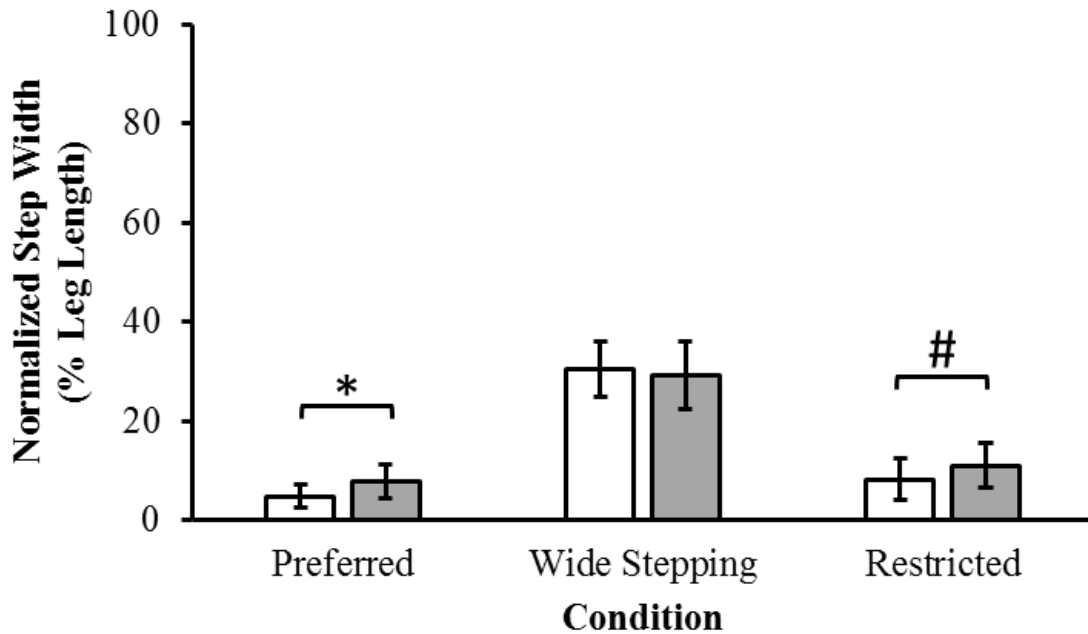
Figure 5-8: Mean(SD) values for the normalized step width. Restricted refers to the restricted arm movement condition. * $p \leq 0.05$; # $0.05 < p \leq 0.10$.

The initial tether load was influenced by a condition x age group interaction ($F(2,68)=3.78$; $p=0.028$). Post-hoc analysis showed that tether load during the restricted arm movement condition was higher in the older adults, vs. the young adults ($p=0.023$) (Table 5-8). The pre-release AP stability margin was also influenced by a condition x age group interaction ($F(2,68)=5.21$; $p=0.008$), where during the restricted arm movement condition, the older adults tended to exhibit a smaller AP stability margin before release, compared to the young adults ($p=0.074$) (Table 5-8).

Table 5-8: Mean(SD) values for the initial conditions calculated prior to the point of cable-release.

Initial Condition Variable	Young Adults			Older Adults		
	First 10 Trials	Wide Stepping	Arm Restrict.	First 10 Trials	Wide Stepping	Arm Restrict.
Tether Load (%BW)	10.4(0.5)	10.4(0.5)	10.0(0.6)[a]	10.5(0.7)	10.5(1.1)	10.6(0.9)[a]
R Vert. Force (% Total)	49.9(0.4)	49.9(0.4)	50.0(0.4)	49.8(0.9)	49.9(1.0)	50.1(1.1)
L Vert. Force (% Total)	50.1(0.4)	50.1(0.4)	50.0(0.4)	50.2(0.9)	50.1(1.0)	49.9(1.1)
AP Stab Marg[#] (mm)	-14.9(24.0)	-9.2(28.0)	-6.1(24.2)[b]	-18.9(26.7)	-18.8(28.2)	-22.5(29.1)[b]
ML Stab Marg[#] (mm)	202.6(8.4)	205.2(11.4)	203.9(9.2)	205.0(9.3)	204.5(9.6)	203.8(10.2)
AP COM Start Pos. (mm)	449.9(26.4)	443.4(31.3)	439.9(25.9)	450.3(29.7)	453.2(32.9)	453.2(30.6)
ML COM Start Pos. (mm)	8.3(4.3)	6.8(10.9)	7.8(4.6)	7.6(6.1)	9.1(6.3)	9.4(6.1)

[a] = $p\leq0.05$; [b]= $p\leq0.10$; BW = body weight; R = right; L = left; Stab Marg = stability margin; Pos. = position; COM = center of mass.
[#]Note: stability margins reflect the period prior to cable-release.

5.5   *Discussion*

   The objective of this study was to assess how wide stepping and restricted arm movement influence balance control during the landing phase of forward reactive stepping. Not surprisingly, the wide stepping condition resulted in the largest ML body movement after FC, regardless of age group. However, wide stepping also resulted in the largest AP body movement after FC. If wide stepping is a proactive strategy used by older adults to preserve ML stability, inducing greater AP body movement would be counterproductive to such a goal. Second, despite limited AP influence, restricted arm movement resulted in larger ML body movement after FC, compared to the preferred stepping condition. This suggests that arm movement may be important for ML control during reactive stepping, and is in-line with past research suggesting that compensatory arm movement is particularly important in older adults

   Regarding the first hypothesis, wide stepping was expected to result in the largest and most variable peak AP and ML COM displacement after FC. In terms of the magnitude of the displacement, this hypothesis was true as wide stepping did result in the largest COM displacement in the AP and ML directions. As many studies have reported that older adults take wider steps (Dean, et al., 2007; McIlroy & Maki, 1996; Rogers, et al., 2001; Schrager, et al., 2008; Schulz, et al., 2005; Troy, et al., 2008), possibly as part of a strategy to preserve stability in both directions (Schulz, et al., 2005), greater AP displacement may be counter productive to this goal. Accordingly, for both age groups, step length was also the largest during the wide stepping condition (Table 5-6), suggesting participants were able to compensate accordingly. In terms of the variability of the AP COM displacement after FC, restricted arm movement resulted in the least variable displacement (vs. preferred and wide stepping), but preferred and wide stepping were not different from each other. In the ML direction, the wide stepping condition resulted in the greatest variability in the peak COM displacement after FC, in line with our hypothesis. As this was a main effect, both young and older adults exhibited this increased variability. Due to the large step width required during this condition, perhaps participants were 'afforded' greater freedom in the control of their COM after FC (trial-to-trial), due to a potentially larger stability margin.

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To gain further insight into the wide stepping condition, the ML stability margin at the peak COM position after FC was also examined. Although the peak ML COM displacement after FC was largest during the wide stepping condition, step width was also largest during this condition. This resulted in the largest ML stability margin during the wide stepping condition. Previously, researchers have postulated that COM movement after FC was associated with (1) challenges in reactive control during the landing phase; or (2) a strategy to simplify reactive control in the event that additional forward steps were required (Singer, et al., 2013). Research on voluntary stepping in young adults showed that COM displacement after FC was largest during a wide-stepping condition; however, when COM displacement was referenced to the base-of-support, the wide-stepping condition was no different than the preferred. To the authors, this suggested a (proactive) attempt to scale the peak ML COM displacement to step width, to take advantage of the passive dynamics of the response (Singer, et al., 2012). In the current study, the large ML COM displacement in the wide stepping condition could suggest that participants were also scaling their ML COM displacement to the large step width. Interestingly, in subsequent work, the authors concluded that dyscontrol (not a proactive strategy) was more likely, due in part to the increased trial-to-trial variability (in COM displacement after FC) in older adults during voluntary and reactive stepping (Singer, et al., 2013, 2016). The current results do not provide direct support for the dyscontrol hypothesis. Instead, the wide stepping movement was likely responsible (in part) for the large ML COM displacement after FC. The lateral stepping motion, which involved a large degree of hip abduction, may have caused the body to move or rotate laterally, due to the creation of a large frontal plane moment vs. the other two conditions. As mentioned above, this is in agreement with the passive dynamics hypothesis from previous research (Singer, et al., 2012).

Next, it was hypothesized that restricted arm movement would result in larger and more variable peak COM displacement after FC compared to the preferred condition. For the AP COM displacement, there was no difference between the restricted arm movement and preferred stepping conditions. However, ML COM displacement was on average 12.3 mm larger (across both groups) in the restricted arm movement condition vs. the preferred stepping condition. However, the restricted arm movement condition

did not result in more variability, and in the AP direction, restricted arm movement actually resulted in the lowest trial-to-trial variability (Table 5-3). These results suggest that the arms may be more important for frontal-plane control, to help stabilize the body during the landing phase. Previous research on the landing phase did not restrict arm movement during any of the trials (Singer, 2012; Singer, et al., 2016). However, the authors suggested that arm movement during the stepping phase, and the corresponding increase in the moment of inertia about the anteroposterior axis, could moderate frontal-plane angular acceleration caused by the gravitational force, which may simplify stability control during the landing phase (Singer, 2012). Interestingly, in the ML direction, the stability margin was larger during the restricted arm movement, compared to the preferred stepping condition, which suggests that there was actually greater stability after FC with no arm movement. Instead, this was likely due to participants taking wider steps in the restricted arm movement condition (vs the preferred condition), to account for a real (or perceived) instability, accentuated by the reduced arm movement. Arm movement did not cause changes in step length, compared to the preferred stepping condition (Table 5-6).

Third, it was hypothesized that the older adults would exhibit a larger, more variable peak AP and ML COM displacement after FC, compared to the young adults. However, the between-group differences would be largest during the wide stepping and restricted arm movement conditions. In the AP direction, this was not true for COM displacement, as an age main effect was observed (Table 5-1). The variability of the AP COM displacement was also unaffected by age. In the ML direction, the older adults exhibited a larger peak ML COM displacement after FC in the preferred condition, which agrees with existing literature (Singer, et al., 2013, 2016) and tended to exhibit a larger ML COM displacement in the restricted arm movement condition. It has been suggested that older adults are reliant on their arms for balance recovery after a perturbation (Maki, et al., 2000), despite deficits in the speed (Allum, et al., 2002; Maki, et al., 2001; Mansfield & Maki, 2009; Robinovitch et al., 2005; Weaver, et al., 2012), or magnitude (Allum, et al., 2002) of these responses when compared to young adults. The results of the current study add to this notion, particularly for ML COM control during the landing phase of reactive stepping. Researchers have suggested that future studies should focus on training protective arm reactions for fall-related injury

prevention (Choi et al., 2015; Feldman & Robinovitch, 2007; Schonnop, et al., 2013), but perhaps training arm movements for balance recovery may also be fruitful. Nonetheless, older adults may have compensated for the trend of increased ML COM displacement by taking wider normalized steps during the restricted arm movement condition, compared to the young adults (10.97(4.49) % vs. 8.15(4.08) %; p=0.057). The lack of age-related differences in peak ML COM displacement during the wide stepping condition is in-line with previous research where older adults were not more unstable (i.e., a larger margin of stability at heel-strike) than young adults when taking progressively larger lateral steps. Instead the older adults may have compensated by generating a larger peak hip abductor moment (of the stepping limb) which may reflect greater muscular effort by older adults to reduce the likelihood of becoming unstable (Hurt & Grabiner, 2015).

As with Studies 1 and 2, Study 3 was primarily limited by the uni-directional nature of the tether-release perturbation. This coupled with the fact that participants were instructed to respond with a single forward step suggests that participants may have been able to pre-plan aspects of their reactive stepping responses. Accordingly, all results must be interpreted with these limitations in mind. Study 3 was also limited by the fact that the older adults were all community-dwelling, and active. Therefore, future studies should aim to investigate the same stepping conditions in groups at a higher risk of falling, such as frail older adults and those with neurological conditions. Next, only one wide stepping condition was examined, which was based on the physical dimensions of the force-plate which participants stepped onto. It is also unclear how the 'wide step' relates to real-world conditions, as it was chosen to be wide enough to challenge participant capabilities. Further, during the restricted arm movement condition, no physical restriction was placed on arm movement; however, the researchers were vigilant in their observation of whether participants kept their arms by their sides during these trials. Limitations aside, this was the first study to assess how wide stepping and restricted arm movement influence balance control during reactive stepping, specifically during the landing phase.

In conclusion, wide stepping resulted in the largest ML body movement after FC, in both age groups. Interestingly, wide stepping also resulted in the largest AP body movement after FC. However,

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ML stability margins were also the largest during the wide stepping condition, suggesting that taking a wide step may be a positive strategy for improving ML stability during the landing phase of reactive stepping. During the restricted arm movement condition, older adults tended to exhibit larger ML COM movement after FC, which they may have accommodated for by taking wider (normalized) steps. Looking ahead, future research (Chapter 6) must expand to studying the landing phase during multi-step scenarios, evoked using larger perturbations. Evoking and studying multi-step responses may be more relatable to real-life losses of balance where reactive stepping is required, such as tripping while walking.

6. STUDY FOUR – QUANTIFYING BALANCE CONTROL AFTER FOOT-CONTACT DURING TWO-STEP RESPONSES

6.1 Chapter Overview

While the landing phase of voluntary and reactive single-step responses has recently become a focus of researchers, multi-step responses are often used in real-world situations. The primary purpose of this study was to determine if single-step responses are similar to those observed during multi-step scenarios. This was done by comparing the first step of one and two-step responses and the first and second step of two-step responses. The secondary purpose was to assess effects of asymmetrical body weight while leaning which is relevant for individuals with loading asymmetries. Eighteen young adults participated in one and two-step conditions, where the initial lean was symmetrical or asymmetrical. Peak center of mass (COM) and extrapolated COM (xCOM) displacement after foot-contact were calculated in the anterior-posterior (AP) and medio-lateral (ML) directions. Compared to one-step responses, two-steps resulted in AP and ML COM displacements during the first step which were larger and smaller, respectively ($p<0.001$-$p=0.009$). With asymmetrical loading, first step lengths were larger during the one-step condition ($p=0.002$), while first step width was reduced over both stepping tasks with asymmetrical loading ($p<0.001$). Peak AP xCOM displacement after foot-contact was larger in the second, compared to the first step with asymmetrical loading ($p=0.023$), yet the first step resulted in greater ML xCOM displacement vs. the second step ($p<0.001$), regardless of loading. First step width was narrower than the second step with asymmetrical loading ($p=0.006$). Differences between one and two-step responses suggest that single-step responses should not be used to infer balance control during multi-step scenarios.

6.2 Introduction

The most common types of perturbations are voluntarily in nature. One example, gait termination, is particularly difficult because of the need for the central nervous system to predict the future and final position of the body's center of mass (COM) (Winter, 1995). As such, older adults terminate their gait less frequently using just one-step in comparison to young adults (Menant, et al., 2009; Tirosh &

Sparrow, 2004, 2005), with longer mean stopping times (Tirosh & Sparrow, 2004). It is unclear if this need for multiple steps in older adults is real or "perceived" as 86% of the older adults' two-step responses occurred within the predicted stability region prior to them taking the second step (Tirosh & Sparrow, 2004). Research on the landing phase of reactive stepping when participants use two-steps has important implications for activities such as rapid gait termination or tripping during walking which is common in frail older adults (Robinovitch, et al., 2013), and it may further inform researchers regarding the role of body movement after foot-contact. However, recent studies have focused primarily on the landing phase during only single-step responses (King, et al., 2012; Serrao, et al., 2013; Singer, et al., 2014; Singer, et al., 2012, 2013, 2016), therefore, comparing one and two-step responses will inform whether body movement and foot-placement during single-step responses are applicable to multi-step scenarios.

In the process of quantifying landing phase control during two-step responses, it is also necessary to consider the effects of body weight asymmetry. During perturbations such as tripping while walking and voluntary stepping, the support-leg is typically loaded to a greater degree than the step-leg. Interestingly, anticipatory postural adjustments (APAs) are normally either absent or diminished in magnitude and effectiveness during reactive stepping (McIlroy & Maki, 1999; Rogers, et al., 2001). While this may help to achieve a rapid response time, it may also have a negative effect on medio-lateral (ML) stability (McIlroy & Maki, 1999; Rogers, et al., 2001). Conversely, voluntary steps almost always occur with an APA preceding stepping (Brunt, et al., 1991). Unloading the step-leg during reactive stepping (evoked from standing) could allow for a faster toe-off time, as less time would be needed to unload the step-leg, which might also influence COM kinematics during the landing phase, and foot-placement. Previous research has assessed the effect of asymmetrical loading on reactive step placement and spatio-temporal step characteristics (Lakhani et al., 2011); however, participants did not step with the unloaded leg in 100% of trials, making it difficult to determine how body weight asymmetry influenced step length and width. Such research is relevant for real-world perturbations such as tripping during gait,

or for individuals with chronic stroke, where up to 56% of these individuals exhibit gait asymmetry (Patterson et al., 2008).

As such, the overall goal of this study was to determine if single-step responses are similar to those observed during multi-step scenarios, in terms of body movement after foot-contact (FC) and foot-placement (step length and width). Comparing landing phase control during one and two-step responses will also provide further insight into the role of body movement after FC. Therefore, the specific objectives were to: 1) compare balance control and foot-placement during one-step and two-step responses and to assess the effect of asymmetrical loading, using only the first step; 2) compare balance control and foot-placement between the first and second steps of two-step responses, and examine the effects of asymmetrical loading during two-step responses; and 3) determine if balance control and foot-placement of the first and second steps is related. The first step of both one-step and two-step responses was compared to the second step of two-step responses to provide further insight into whether balance control during multi-step responses can be inferred from one-step responses.

When comparing the first step, it was hypothesized that: 1a) two-step responses would result in a larger first step peak COM displacement after FC, compared to one-step responses, however step length and width would be larger during the one-step responses. 1b) Additionally, it was hypothesized that asymmetrical loading (with an increased load on the support-leg) would result in a smaller peak COM displacement after FC of the first step, along with a longer and wider first step, compared to symmetrical loading (for both one and two-step responses). Second, during the two-step responses, 2a) the second step (vs. the first step) would occur with a smaller peak extrapolated COM (xCOM) displacement after FC, and be longer and wider. 2b) Asymmetrical loading (with an increased load on the support-leg) would result in a smaller peak xCOM displacement after FC during the first step, but a larger peak xCOM displacement after FC during the second step. Asymmetrical loading (with an increased load on the support-leg) would also result in a longer and wider first step. 3) Lastly, it was hypothesized that: 3a) the peak xCOM displacement after FC of the first and second step would moderately-to-strongly positively (anterior-posterior) and negatively (medial-lateral) correlate. 3b) Step length during the first and second

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step would moderately-to-strongly positively correlate, while step width would moderately-to-strongly negatively correlate. 3c) First step metrics calculated from the one-step only responses would not be correlated to the second step during two-step responses. Note that the COM was used for the analysis of the first step, in order to allow for direct comparison with previous research which has focused on one-step responses (Singer, 2012; Singer, et al., 2016). The xCOM was used for the comparison of the first and second steps, because of the need to consider COM velocity during multi-step scenarios. For example, in the anterior-posterior direction, one must redirect the COM velocity from one pendular arc to the next during the transition between steps (Donelan et al., 2002). In the medio-lateral direction, one must also redirect the COM velocity during the transition between single stance phases (Donelan et al., 2001).

6.3   Methods

6.3.1   Experimental Protocol

Eighteen young adults participated in Study 4 (age: 25.1(2.7) y; height: 1.7(0.1) m; mass: 74.7(14.4) kg; 8 females). All were right leg dominant, defined as the leg they indicated they would kick a soccer ball with. Ethics clearance was obtained from the University of Waterloo Human Research Ethics Committee, and all participants provided informed consent prior to participating.

Twelve cameras (Optotrak Certus, Northern Digital Incorporated, Waterloo, Ontario, Canada) were used to collect kinematic data at 64Hz, while three force-plates (BP 5050 (x2), Bertec, Columbus, Ohio, USA; and OR6-7 (x1), Advanced Mechanical Technology Inc., (AMTI) Watertown, MA, USA) were sampled at 2048 Hz. The two Bertec force-plates were arranged side-by-side (Figure 6-1 and Figure 6-2). The tether supporting each participant's body weight was located in-line with a load cell (MLP-300-CO, Transducer Techniques, Temecula, CA) which was sampled at 2048 Hz and rated for up to 136 kg. The tether was connected to a metal frame via an electromagnet (AEC Magnetics, Cincinnati, OH, USA).
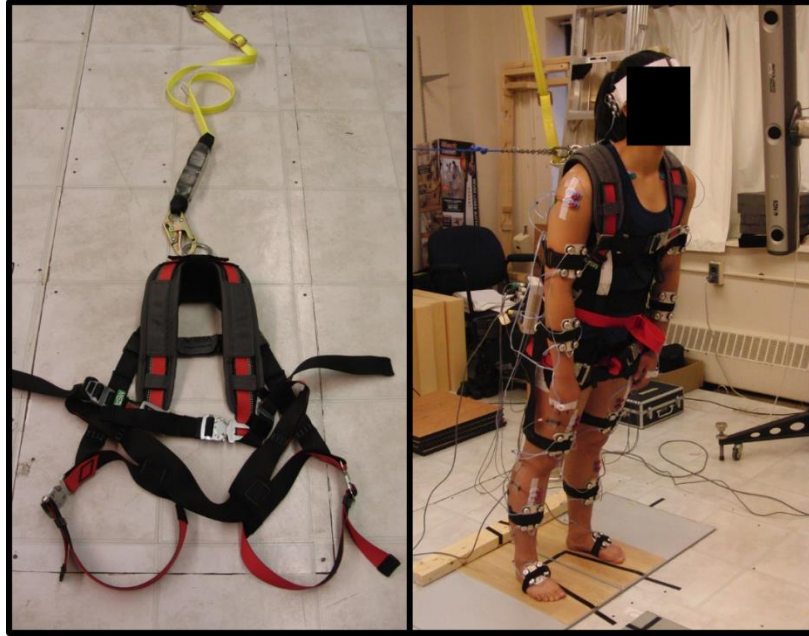
Figure 6-1: On the left is the safety harness which participants wore, along with the safety-tether which was connected to the ceiling. On the right, a participant is depicted wearing the harness, along with the safety tether. The two Bertec force-plates are also depicted.
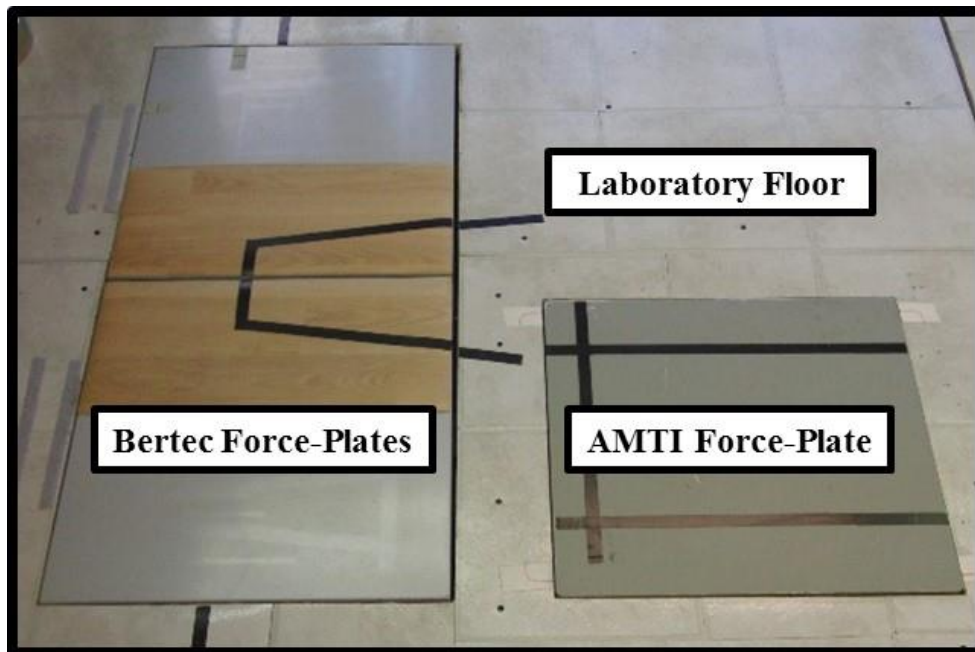
Figure 6-2: The layout of the three force-plates is depicted. The AMTI force-plate is where all steps with the right leg landed, and the laboratory floor is where all steps with the left foot landed.

In the current study, ankle muscle electromyography (EMG) and full body kinematics were collected. Electromyographic data was sampled at 2048 Hz using a differential amplifier, with a hardware band-pass filter of 10-1000 Hz, a common mode rejection ratio of 115dB at 60 Hz (Bortec Biomedical, Calgary, AB) and disposable, self-adhesive Ag/Ag-Cl electrodes which were placed bilaterally on the tibialis anterior (TA) and medial gastrocnemius (MG). All motion capture data and analog-to-digital converted signals were synchronized using First Principles software (Northern Digital Incorporated, Waterloo, Ontario, Canada). The analog-to-digital converter included a 16-bit card. Kinematics were measured using a whole-body marker set. Rigid clusters of four markers were placed on the locations depicted in Appendix 2. Additionally, using a digitizing probe (Northern Digital Incorporated, Waterloo, Ontario, Canada) "imaginary" markers were digitized bilaterally at anatomically relevant locations (Appendix 2). The laboratory global coordinate system was defined in accordance with ISB recommendations (Wu & Cavanagh, 1995).

Two quiet standing trials were first collected. The first had participants adopt a standardized foot position, based on existing research (0.17 m between heel centers and an angle of 14º between the long axes of the feet) (McIlroy & Maki, 1997), with their arms at their sides, while looking straight ahead. The second trial required participants to start in the standardized foot position, take a single forward step onto the AMTI force-plate with their right (dominant) leg, and hold the final forward-stance position (Singer, 2012). Both trials were 60 seconds in duration (Carpenter, et al., 2001). After which, all participants completed practice trials, single-step trials and two-step trials.

To provide the external postural perturbations, a tether-release paradigm was used (Hsiao-Wecksler, 2008) (Figure 6-1). Participants started in an initial forward-lean position pertaining to 12-13% of their body weight, which was monitored in real-time and kept to within ± 1% body weight trial-to-trial (Graham, et al., 2015). Each participant's initial stance width was standardized as described above. Participants initially stood with their feet on separate force-plates, which allowed for the center of pressure position of each foot and the body weight supported by each leg (Newtons) to be monitored prior

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to each tether-release perturbation (trial-to-trial) using a real-time LabVIEW feedback routine (National Instruments Corporation, Austin, TX).

The EMG activity recorded bi-laterally from the TA and MG was used to ensure that the level of pre-perturbation EMG activity during the tether-release trials would not exceed the maximum from the feet side-by-side quiet standing trial (Singer, 2012; Singer, et al., 2016). This was done using a LabVIEW feedback routine (National Instruments Corporation, Austin, TX). Verbal encouragement was provided emphasizing participants to, "allow for the tether to fully support their body weight". The tether was released at random intervals after the initial conditions were met. A minimum of one second was always captured from when the participant met the initial conditions and the tether was released.

Upon completion of the kinematic and EMG setup, reactive stepping was evoked during multiple tether-release conditions, each with separate instructions. Five practice trials were always completed first. Participants were instructed to do what felt "natural", as the goal was to familiarize participants with the perturbation magnitude, but not a specific reactive stepping response. During the single-step trials, participants responded to each tether-release perturbation with one-step using their right leg, directly onto the AMTI force-plate. During the two-step trials, participants responded to each tether-release perturbation using two-steps. The first step was with their right leg, onto the AMTI force-plate, while the second step was taken with the left leg onto the laboratory floor beside the AMTI force-plate (Figure 6-2). No instruction was provided regarding where the left foot should land, as long as it did not touch the AMTI force-plate, as the researcher wished to avoid further movement constraints. Participants started all trials with their arms at their sides.

During each of the two blocks described above, participants completed: 1) five trials with their body weight evenly (50%/50%) distributed between their legs while leaning, and 2) five trials with 60% of their body weight over the left (initial non-stepping) leg and 40% of their body weight over the right (initial stepping) leg (Lakhani, et al., 2011). Originally, a 70%/30% ratio (Lakhani, et al., 2011) was tested in pilot sessions. However, the 70% support-leg load, coupled with the 12-13% lean magnitude, was too difficult to maintain pre-perturbation and also resulted in participants often missing the force-

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plate with their reactive step. The order in which the four experimental conditions (Table 6-1) were presented was block randomized between participants. Regardless of experimental condition, participants were instructed to maintain their final position for approximately 10 seconds once they regained their stability (Singer, 2012).

Table 6-1: Summary of the experimental conditions.

EXPERIMENTAL CONDITION	DEPENDENT VARIABLES
1) One-step, symmetrical body weight 2) One-step, asymmetrical body weight	1) Peak AP COM displacement after foot-contact 2) Peak ML COM displacement after foot-contact 3) Step length 4) Step width
3) Two-steps, symmetrical body weight 4) Two-steps, asymmetrical body weight	1) Peak AP xCOM displacement after foot-contact 2) Peak ML xCOM displacement after foot-contact 3) Step length 4) Step width

AP = anterior-posterior; ML = medio-lateral; COM = center of mass; xCOM = extrapolated center of mass.

6.3.2   *Data Analysis*

Force-plate data was low-pass filtered using a 2nd order, dual-pass Butterworth filter with a cut-off frequency of 50 Hz (Singer, et al., 2016). Toe-off was defined as the point when the vertical force under the right and/or left leg fell below 10 N (Sparrow & Tirosh, 2003), while FC was defined as the time point of the minimum vertical velocity (of the foot COM) after toe-off of each step (O'Connor et al., 2007). Cable-release was calculated using the data from the load cell located in-series with the tether. This data was low-pass filtered using a 2nd order, dual-pass Butterworth filter, with a cut-off frequency of 3 Hz (Wright, et al., 2014). Cable-release was defined in accordance with previous research (Graham, et al., 2015), as a 20% reduction in force measured using the load cell located in-series with the tether.

All kinematic data was low-pass filtered using a 2nd order, dual-pass, Butterworth filter with a cut-off frequency of 6 Hz (Graham, et al., 2015; Singer, et al., 2016). An estimate of the whole body center of mass (COM) was calculated using the filtered kinematic data and the anthropometric tables of de Leva (1996). To determine the hip and shoulder joint centers, the methods of Weinhandl and O'Connor (2010) and Nussbaum and Zhang (2000), respectively, were used. Next, the position of the COM, in the anterior-posterior (AP) and medio-lateral (ML) directions, was calculated at the following time points: 1) toe-off, 2) FC and, 3) the peak COM position after FC. This was done for each step. During the two-step responses, the peak COM position after FC of the first step was calculated up until the instant of toe-off of the second step. In each instance, to calculate COM displacement, the COM was referenced to the mean starting COM value, which was calculated as the average from the start of the trial (frame 1) to one frame before cable-release. The xCOM was also calculated, in accordance with Hof, Gazendam and Sinke (2005), at the same time points as the COM. Only the xCOM was used for the comparison of the two-step responses because stance leg transitions require mechanical work to redirect the ML COM velocity from one leg to the next (Donelan, et al., 2001). The xCOM of both the one and two-step responses was also used for the correlational analyses detailed below in the Statistical Analyses section. For both the COM and xCOM, the direction of the peak ML position after FC was determined based on the stepping leg (i.e., peak right (+Z) position for right steps, and peak left (-Z) position for left steps). Values calculated during the two-step responses with the left leg were mirrored to allow for statistical comparison to the right-legged steps. Step length and width were calculated using the COM of the right and left feet. For each right or left step, the difference in the position of the foot COM between FC and cable-release was calculated as the step length (AP) and step width (ML), respectively.

To examine potential confounding factors related to participant position prior to cable-release, multiple variables were calculated. The mean tether-load (during the lean) and the mean vertical ground reaction forces under the right and left feet were each calculated. Anterior-posterior and ML stability margins were also calculated by subtracting the COM position from the tip of the big toe and the 5th metatarsal, in the AP and ML directions, respectively. Stability margins were calculated with respect to

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the right and left feet, due to the two-step conditions. For each trial, mean initial condition metrics were calculated from the start of the trial to one frame prior to cable-release.

6.3.3   Statistical Analyses

Primary Analyses: To compare balance control after FC during the first step of one and two-step responses (hypotheses 1a – 1b), a 2 x 2 repeated measures ANOVA was used with stepping task (one-step vs. two-steps) and body weight symmetry (symmetrical vs. asymmetrical) as factors. Second, to examine balance control during two-step responses (hypotheses 2a – 2b), a 2 x 2 repeated measures ANOVA was also used with step number (first vs. second), and body weight symmetry (symmetrical vs. asymmetrical) as factors. These analyses were conducted using the: 1) peak COM\xCOM after FC and, 2) step length and width. Note, COM displacements were compared during the first step analyses (hypotheses 1a – 1b), while xCOM displacements were compared during the two-step responses (hypotheses 2a – 2b). Lastly, to determine if balance control during the first and second steps was correlated (hypothesis 3a – 3c), one-tailed Pearson's correlation coefficients were calculated using the AP and ML peak xCOM displacement after FC, step length and step width from the first and second steps of the two-step responses, as well as the first step values from the one-step only responses. All significant interactions were explored using post-hoc paired-samples t-tests.

Secondary Analyses: A secondary analyses was conducted comparing the initial lean conditions, to ensure that changes in the dependent variables were not caused by different lean positions between experimental conditions. The initial conditions for the first step analyses were compared using repeated-measures ANOVAs as detailed above. To compare the initial conditions during the two-step responses, paired-samples t-tests were used. All statistical analyses were completed using SPSS (v.21, IBM Corporation, New York, USA). Experiment wide statistical significance was set at $p \leq 0.05$. Results of the secondary analyses can be found in Appendix 3 – Supplementary Tables 1 and 2. Additionally, mean(SD) values for the primary analyses can also be found in Appendix 3 – Supplementary Tables 3 to 10.

6.4 Results

A summary of the statistical results can be found in Table 6-2, while mean(SD) values for the first step and first vs. second step analyses are depicted in Figures 6-3 and 6-5, respectively. Lastly, mean time-series COM (first step comparison) and xCOM (first and second step comparison) data can be found in Figures 6-4 and 6-6, respectively.

Table 6-2: Summary of the statistical effects observed for the comparison of only the first step (hypotheses 1a – 1b) and the comparison of the first and second steps (hypotheses 2a – 2b).

Dependent Variable	Effect(s)	ANOVA Output
First Step Comparison – One and Two-Step Responses (Hypotheses 1a -1b)		
AP COM Displacement (mm)	Stepping Task Body Weight Symmetry Interaction	**$F(1,17)=21.24; p<0.001$** $F(1,17)=3.68; p=0.072$ **$F(1,17)=5.55; p=0.031$**
ML COM Displacement (mm)	Stepping Task Body Weight Symmetry Interaction	**$F(1,17)=73.95; p<0.001$** $F(1,17)=1.16; p=0.296$ **$F(1,17)=8.72; p=0.009$**
Step Length (mm)	Step Task Body Weight Symmetry Interaction	**$F(1,17)=10.30; p=0.005$** $F(1,17)=0.06; p=0.810$ **$F(1,17)=7.88; p=0.012$**
Step Width (mm)	Stepping Task Body Weight Symmetry Interaction	**$F(1,17)=8.50; p=0.010$** **$F(1,17)=33.11; p<0.001$** $F(1,17)=1.01; p=0.330$
First and Second Step Comparison – Two-Step Responses (Hypotheses 2a – 2b)		
AP xCOM Displacement (mm)	Step Number Body Weight Symmetry Interaction	$F(1,17)=2.70; p=0.118$ **$F(1,17)=6.00; p=0.025$** **$F(1,17)=5.84; p=0.027$**
ML xCOM Displacement (mm)	Step Number Body Weight Symmetry Interaction	**$F(1,17)=36.83; p<0.001$** $F(1,17)=3.56; p=0.076$ $F(1,17)=3.98; p=0.062$
Step Length (mm)	Step Number Body Weight Symmetry Interaction	**$F(1,17)=17.19; p=0.001$** $F(1,17)=1.41; p=0.251$ **$F(1,17)=7.31; p=0.015$**
Step Width (mm)	Step Number Body Weight Symmetry Interaction	$F(1,17)=1.76; p=0.202$ $F(1,17)=0.003; p=0.956$ **$F(1,17)=48.55; p<0.001$**

AP = anterior-posterior; ML = medio-lateral; COM = center of mass; xCOM = extrapolated center of mass.

First Step Comparison: The peak COM displacement after FC in the AP direction was influenced by a stepping task (one vs. two-steps) x body weight symmetry interaction ($F(1,17)=5.55$; $p=0.031$), where with symmetrical loading the first step peak AP COM displacement was larger during the two-step (306.8(30.9) mm) compared to the one-step condition (274.2(32.3) mm) ($p<0.001$). During the asymmetrical body weight condition, the two-step condition (308.3(29.8) mm) also resulted in a larger first step peak AP COM displacement compared to the one-step condition (288.3(28.2) mm) ($p=0.009$). In the ML direction, the peak COM displacement after FC was also influenced by a stepping task x body weight symmetry interaction ($F(1,17)=8.72$; $p=0.009$). With symmetrical loading, the first step peak ML COM displacement was lager during the one-step (83.3(13.3) mm) compared to the two-step condition (60.4(12.0) mm) ($p<0.001$). During the asymmetrical body weight condition, the one-step condition (90.7(15.8) mm) also resulted in a larger first step compared to the two-step condition (56.5(15.7) mm) ($p<0.001$) (Figures 6-3 and 6-4).
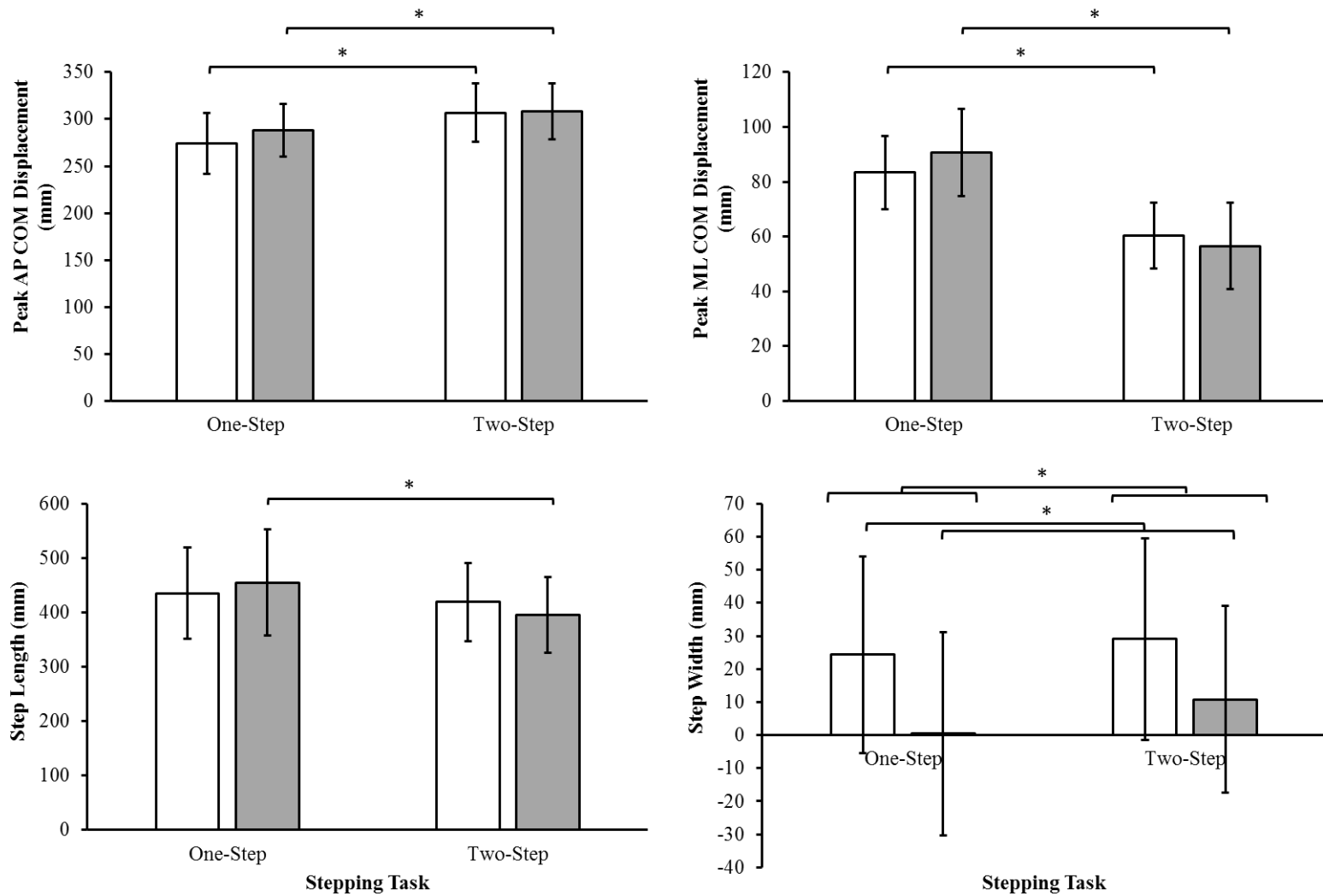
Figure 6-3: First step comparison mean(SD) values for: 1) peak AP COM displacement after FC (top left); 2) peak ML COM displacement after FC (top right); 3) step length (bottom left); and 4) step width (bottom right). * $p \leq 0.05$. AP = anterior-posterior. ML = medio-lateral. COM = center of mass.
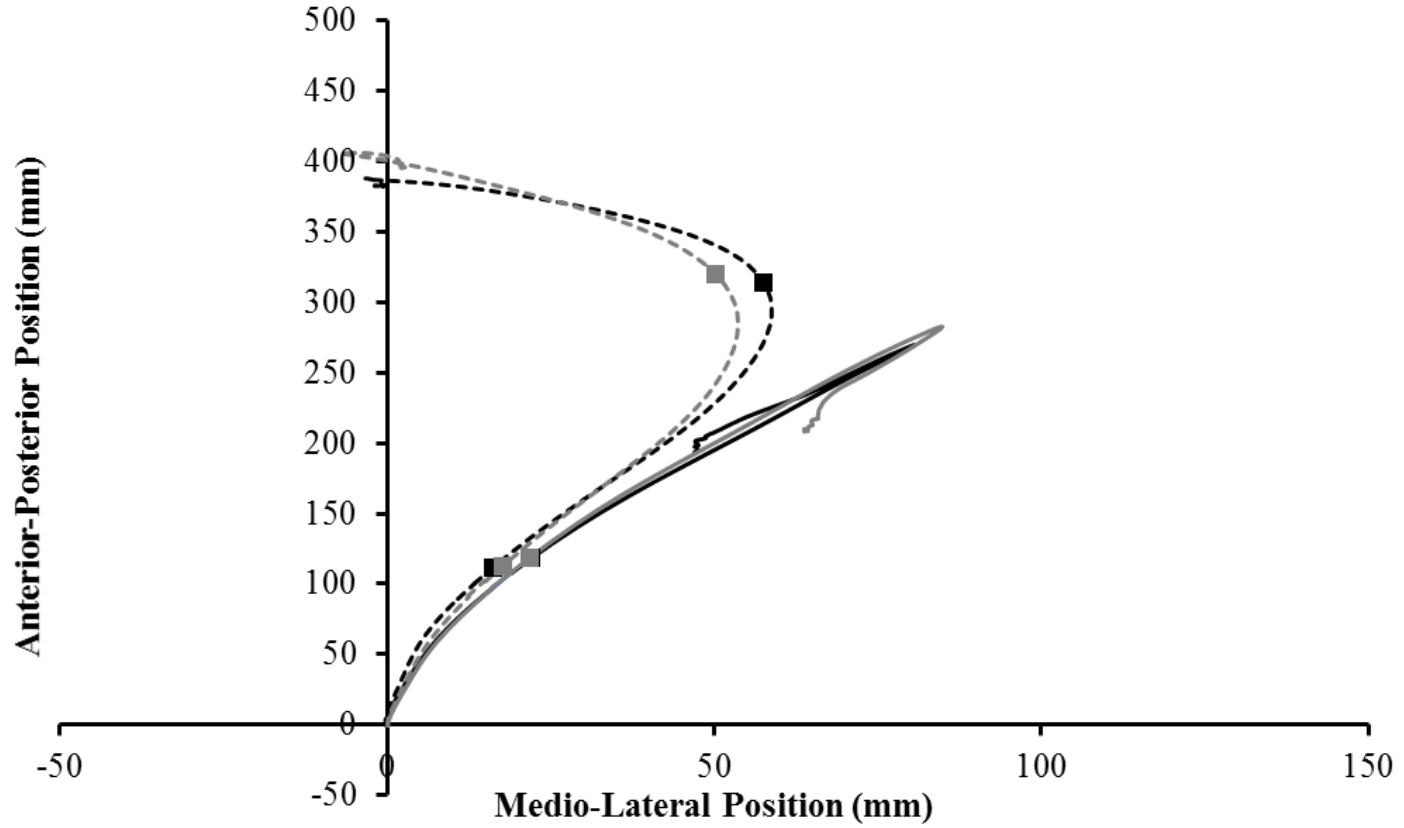
Figure 6-4: Time-series center of mass trajectories for the one and two-step conditions, with symmetrical and asymmetrical loading. The squares represent the center of mass position at foot-contact (for both steps during the two-step condition). Solid lines represent the one-step condition, while dashed lines represent the two-step condition. Black lines represent symmetrical loading, while gray lines represent asymmetrical loading.

Regarding step length, there was a stepping task x body weight symmetry interaction ($F(1,17)=7.88$; $p=0.012$). With symmetrical loading, first step length was not different between the one and two-step conditions ($p=0.170$). During the asymmetrical body weight condition, first step length was larger during the one-step condition (455.0(97.7) mm) compared to the two-step condition (395.1(69.6) mm) ($p=0.002$). Step width was influenced by both stepping task ($F(1,17)=8.50$; $p=0.010$) and body weight symmetry ($F(1,17)=33.11$; $p<0.001$) main effects. First steps were wider during the two-step responses (19.9(30.5) mm) compared to the one-step responses (12.4(32.2) mm). Second, the first step was wider during the symmetrical (26.7(29.8) mm) vs. asymmetrical (5.6(29.6) mm) body weight condition, regardless of step condition.

First and Second Step Comparison (xCOM): The peak AP xCOM after FC was influenced by a step number x body weight symmetry interaction ($F(1,17)=5.84$; $p=0.027$). Post-hoc analyses revealed that AP xCOM displacement during the first and second steps did not differ with symmetrical loading ($p=0.602$); however, with asymmetrical loading, the first (448.0(38.7) mm) and second steps (457.6(52.1) mm) were significantly different ($p=0.023$). In the ML direction, the peak xCOM after FC was influenced by a step number main effect ($F(1,17)=36.83$; $p<0.001$) and a trend of body weight symmetry ($F(1,17)=3.56$; $p=0.076$). Peak ML xCOM displacement was larger during the first step (106.1(18.4) mm) compared to the second step (43.5(36.8) mm). Asymmetrical body weight tended to result in a larger ML xCOM peak displacement after FC (77.3(40.3) mm) compared to the symmetrical body weight condition. (72.2(45.5) mm) (Figures 6-5 and 6-6).

Figure 6-5: First vs. second step comparison mean(SD) values for: 1) peak AP xCOM displacement after FC (top left); 2) peak ML xCOM displacement after FC (top right); 3) step length (bottom left); and 4) step width (bottom right). * $p \leq 0.05$. AP = anterior-posterior. ML = medio-lateral. xCOM = extrapolated center of mass.
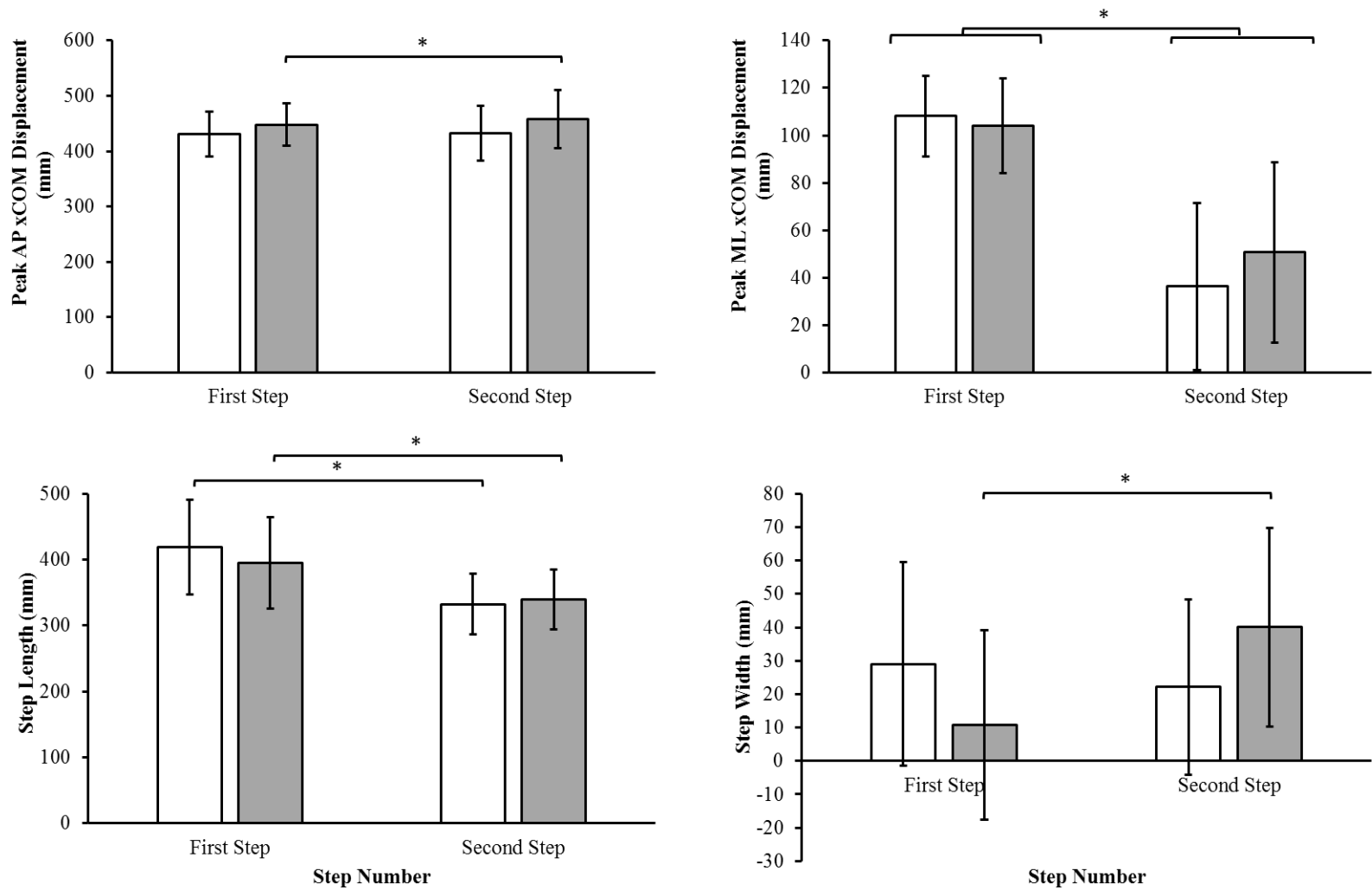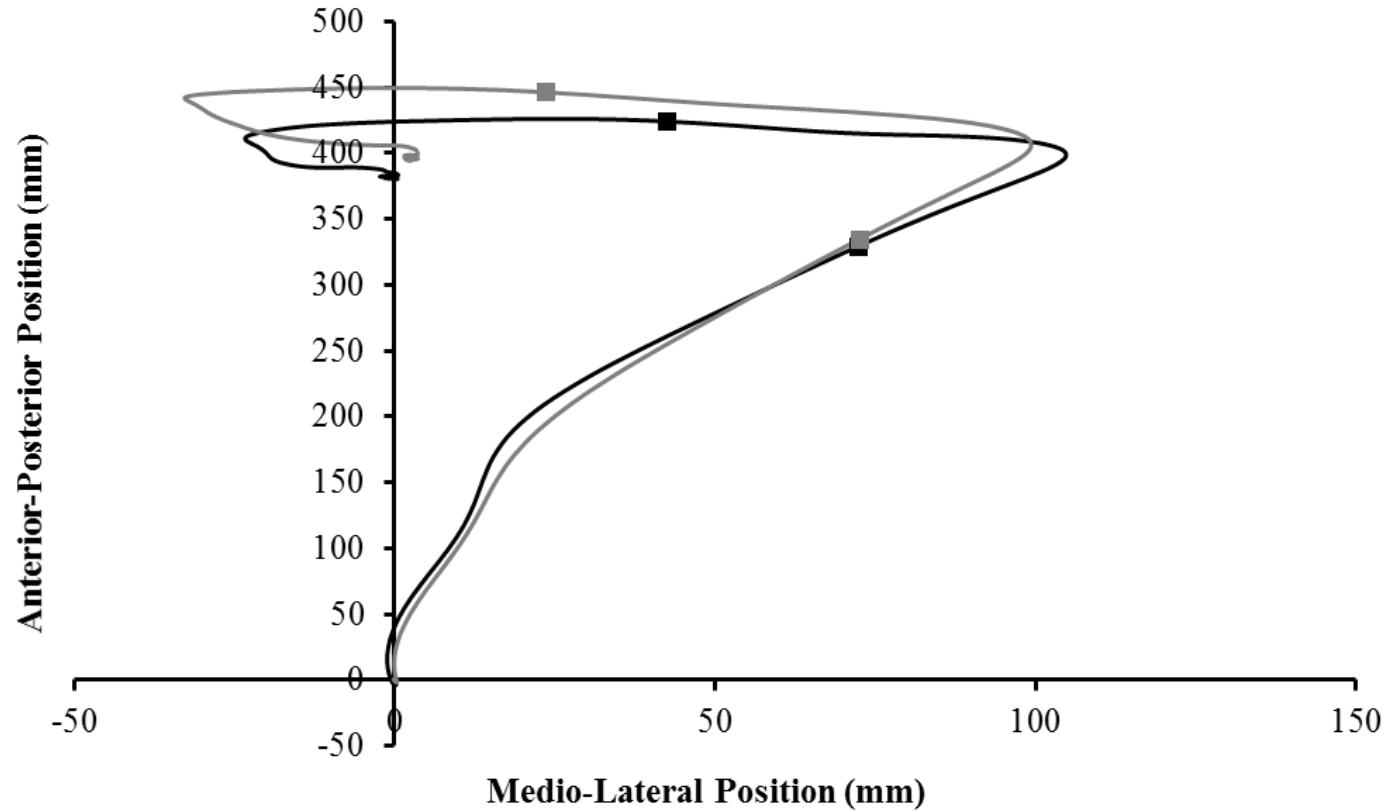
Figure 6-6: Time-series extrapolated center of mass trajectories for the two-step condition, with symmetrical and asymmetrical loading. The squares represent the extrapolated center of mass position at foot-contact for each step. Black lines represent symmetrical loading, while gray lines represent asymmetrical loading.

Step length was influenced by a step number x body weight symmetry interaction ($F(1,17)=7.31$; $p=0.015$), where the first steps were longer than the second steps during both symmetrical (419.1(72.0) mm vs. 332.3(46.3) mm; $p<0.001$) and asymmetrical loading (395.1(69.6) mm vs. 339.1(45.4) mm; $p=0.008$). Step width was influenced by a step number x body weight symmetry interaction ($F(1,17)=48.55$; $p<0.001$) where the first and second step widths did not differ during the symmetrical body weight condition ($p=0.419$). However, during the asymmetrical body weight condition, the first step (10.8(28.3) mm) was narrower than the second step (40.0(29.7) mm) ($p=0.006$).

Regarding the correlations between the xCOM displacement of the first and second steps, significant correlations were observed for all comparisons in the AP and ML directions ($p\leq0.05$) (Table 6-3, Figure 6-7). When the first step from the one-step only responses was used in the correlations, the only significant correlation was for peak AP xCOM displacement during the symmetrical body weight condition ($r=0.586$; $p=0.005$) (Table 6-3). Regarding the step length and step width correlations between the first and second step of the two-step responses, no significant correlations were observed ($p>0.05$). When the first step from the one-step only responses was used, instead of the first step of the two-step responses, one significant correlation was observed for step length, during the asymmetrical body weight\loading condition ($r=0.402$; $p=0.049$) (Table 6-4).

Table 6-3: Correlation coefficients for the peak AP and ML xCOM displacement after foot-contact, along with statistical significance values.

First Step Data	Body Weight Symmetry	AP xCOM r & *p*-value	ML xCOM r & *p*-value
First Step from Two-Step Responses	**Symmetrical**	.953; *p<0.001*	-.405; *p=0.048*
	Asymmetrical	.977; *p<0.001*	-.500; *p=0.017*
First Step from One-Step Responses	**Symmetrical**	.586; *p=0.005*	-.132; *p=0.300*
	Asymmetrical	.280; *p=0.131*	-.281; *p=0.129*

AP = anterior-posterior; ML = medio-lateral; xCOM = extrapolated center of mass.

Table 6-4: Correlation coefficients for step length and step width, and statistical significance values.

First Step Data	Body Weight Symmetry	Step Length r & p-value	Step Width r & p-value
First Step from Two-Step Responses	Symmetrical	.260; $p=0.149$	.232; $p=0.177$
	Asymmetrical	.106; $p=0.338$	.088; $p=0.364$
First Step from One-Step Responses	Symmetrical	.133; $p=0.300$	-.003; $p=0.495$
	Asymmetrical	.402; ***p=0.049***	.009; $p=0.486$
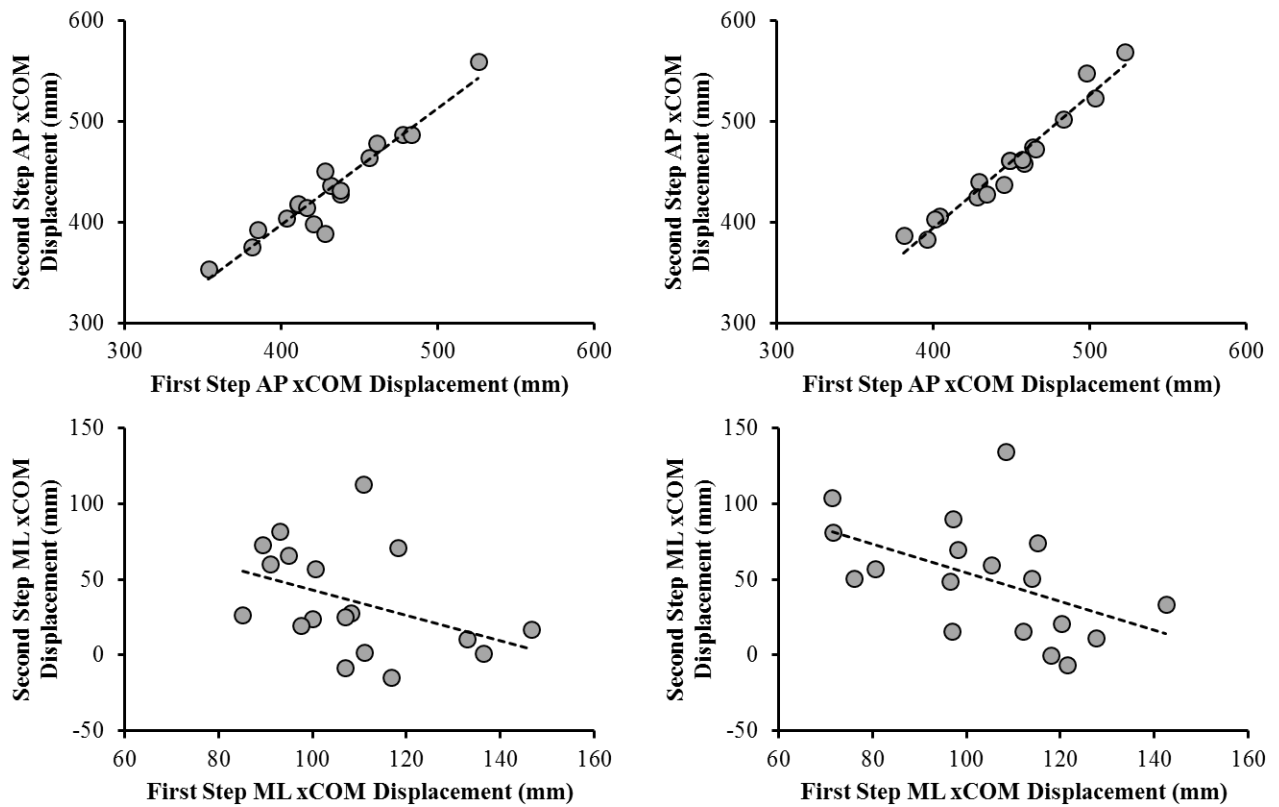
Figure 6-7: Scatter-plots of the correlations between the first and second step peak xCOM displacement after foot-contact, for the anterior-posterior (AP; top-row) and medio-lateral (ML; bottom-row) directions. The symmetrical loading condition is represented by the plots in the left column, while the asymmetrical condition by the plots in the right column. All correlations were significant at $p \leq 0.05$.

6.5   Discussion

This study examined balance control during the landing phase of one and two-step responses. Regarding the first objective to compare balance control during the <u>first step of one and two-step responses</u>, two-steps resulted in AP and ML COM displacements which were larger and smaller, respectively. Asymmetrical loading resulted in larger first step lengths during the one-step condition, while first step width was reduced over both stepping tasks with asymmetrical loading. The second objective was to compare balance control <u>during two-step responses</u> where peak AP xCOM displacement after FC was larger in the second, compared to the first step with asymmetrical loading. However, ML xCOM during the first step was greater compared to the second step regardless of loading. First step width was narrower than the second step with asymmetrical loading. Lastly, during the first and second steps of two-step responses, peak AP xCOM displacements after FC were significantly, positively correlated, while peak ML xCOM displacements after FC were significantly, negatively correlated. Incorporating the first step from the one-step only responses resulted in non-significant correlations, suggesting that control during multi-step responses should not be inferred from responses where only a single-step was used. Specific statistical results will be discussed in greater detail below.

The first objective of this study was to examine peak COM displacement after FC, and foot-placement <u>during the first step of one and two-step responses</u>. The larger peak AP COM displacement after FC, but smaller peak ML COM displacement after FC during the two-step condition, provides insight into the role of COM movement after FC. Previous work suggests that the mechanical work for step-to-step transitions is a large determinant in the metabolic cost of human walking (Donelan, et al., 2002). As such, the larger AP displacement during the two-step condition would potentially reduce the mechanical work needed to move the COM up and over the support limb, where the gait cycle could continue (Singer, et al., 2012). Our results in the AP direction appear to align with the theory that larger AP COM movement in the first step may be part of a multi-step strategy (Singer, et al., 2012). The smaller peak ML COM displacement during the two-step condition suggests that ML COM movement after FC of single-step responses is not purposeful (for a second step) as postulated previously (Singer, et

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al., 2012, 2013) As our sample was a group of young adults, during the one-step condition perhaps participants let their body move more laterally in an exploratory manner as a means of acquiring sensory information (Carpenter et al., 2010; Murnaghan et al., 2013). Interestingly, the stepping task differences were found in peak COM displacement with both symmetrical and asymmetrical loading, suggesting that asymmetrical loading does not have a differential effect on peak AP or ML COM displacement after FC during the first step of one and two-step responses.

As part of comparing the first step, step length and width were also examined. Step length was influenced by a stepping task x body weight symmetry interaction, where symmetrical loading did not influence first step lengths. However, with asymmetrical loading, first step lengths were larger during the one-step condition, compared to the two-step condition. A shorter first step in the two-step condition would be less effortful, and could be beneficial (along with the increased AP COM displacement reported above) for maximizing efficiency if one's goal is to continue forward ambulation. Regarding the step width, it is suggested that there is an increased effort associated with wider steps during gait (mechanical work required to re-direct the COM during the transition between single stance phases) (Donelan, et al., 2001). In the current study, main effects of stepping task and body weight symmetry were observed, where the two-step condition, and symmetrical loading each resulted in wider first steps, compared to the one-step and symmetrical condition, respectively. The main effect of stepping task was likely driven primarily by the task difference within the asymmetrical loading condition (Appendix 3 – Supplementary Table 6), as the mean(SD) step width during the two-step condition was 10.4(17.1) mm larger than the one-step condition. Perhaps during a single-step, young adults may be able to compensate for a narrower step with ground reaction forces of the appropriate size and direction. Overall, asymmetrical loading did result in a narrower first step width, which could be useful when taking a second step, with respect to stability about the stance-leg. For example, given an unchanged ML COM displacement between loading conditions, a narrower step could result in a smaller frontal-plane gravitational moment about the stance-leg after FC. However, the frontal-plane gravitational moment about the stance leg could play a role in minimizing the effort associated with slowing and reversing the ML COM during the first step (making

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the second step easier), as suggested by the wider first step in the two-step (vs. one-step condition), and the large difference between the one and two-step condition with asymmetrical loading (mentioned above).

The second objective of this study was to compare balance control and foot-placement during the first and second steps of two-step responses, while also assessing the effect of body weight symmetry. In the AP direction, step number interacted with body weight symmetry, where the peak AP xCOM displacement after FC did not differ by step number with symmetrical loading. However, with asymmetrical loading, the first (448.0(38.7) mm) and second step (457.6(52.1) mm) were significantly different, where asymmetrical loading may beneficial if the goal is to progress the body forward, such as when taking multiple reactive steps. This result has implications for groups such as stroke patients, as this heterogeneous group has been observed to exhibit asymmetric loading prior to reactive stepping (Mansfield et al., 2012). In the ML direction, the peak xCOM displacement after FC was always smaller during the second, compared to the first step which was in agreement with our hypothesis. A trend of body weight condition was also observed, where asymmetrical loading tended to result in a larger peak ML xCOM displacement after FC. However, this trend of greater peak ML xCOM displacement after FC with asymmetrical loading was likely driven by a larger ML xCOM displacement during the second step, but not the first (Appendix 3 – Supplementary Table 8). This suggests that asymmetrical loading does not differentially influence ML body movement during the first step of multi-step responses.

Similar to the first step comparison, step length and width were also examined during the comparison of one and two-step responses. Step length and width were both influenced by a step number x body weight symmetry interaction. For step length, the first step was longer than the second step, with both symmetrical and asymmetrical loading. This result provides support for our evoked-stepping protocol, and is in agreement with research indicating that gait termination with one's feet parallel rarely occurs during activities of daily living (Hase & Stein, 1998). When examining step width, asymmetrical loading caused the first and second step widths to differ significantly, where the second step was wider. However, symmetrical loading resulted in no difference in step width. This result suggests that

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asymmetrical loading makes it difficult to maintain the first step width (during two-step responses), when compared to symmetrical loading. A similar phenomenon was observed for the comparison of the first step, where asymmetrical loading resulted in narrower mean steps for both stepping tasks.

Generally, our results revealed that balance control during the first and second steps (during two-step responses) are related. In agreement with our final hypothesis, peak xCOM displacement after FC between the first and second steps was very strongly (.80–1.0) (Evans, 1996) positively correlated for all comparisons in the AP direction and moderately (.40-.59) negatively correlated for all comparisons in the ML direction. However, step length and width values between steps were not significantly correlated. These correlations suggest that xCOM movement after FC could be proactive during two-step responses, where larger ML xCOM movement in the first step might aid in taking a more stable second step (as it would reduce the gravitational moment in the frontal-plane causing lateral rotation away from the stance leg, opposite to the notion discussed above). The idea that larger ML COM\xCOM movement after FC may be a strategy to aid in multiple steps has been proposed previously (Singer, et al., 2012). However, as older adults exhibit larger and more variable ML COM movement after FC of single-step responses (Singer, et al., 2013, 2016), it could reflect poor control in older adults, and is unlikely to be proactive when only one-step is required. To this point, when the first step xCOM displacement was calculated from the one-step only responses, the correlations between the first and second step peak xCOM displacement were not as strong in either the AP or ML direction, and only one of four correlations was statistically significant. This suggests that balance control during multi-step scenarios should not be inferred from responses where a single-step was used.

The primary limitation of this study was the fact that the same magnitude perturbation was used to evoke both the one and two-step responses. Although the presentation of these two conditions was block randomized between participants, the fact that all participants could successfully recover using a single step during the one-step condition suggests that the second-step, during the two-step condition, was not necessarily needed for stability. This notion was indirectly highlighted by the fact that the second step lengths were shorter compared to the first step lengths, for both loading conditions (Figure 6-5). However,

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the same magnitude of perturbation ensured that the two conditions could be compared directly. Future studies may wish to evoke one and two-step responses using the same magnitude perturbation, but also evoke both responses at the maximum perturbation magnitudes in which participants can recover from using one and two-steps, respectively. Allowing participants to respond "naturally" to these perturbations, as opposed to using specific task instructions, may also be beneficial for evoking more realistic, less pre-planned responses. Together, these suggestions could help provide further insight into the differences between one and two-step responses, at each participant's maximum capabilities. This study was also limited by the fact that the AMTI force-plate which participants stepped onto constrained where the right foot could be placed. Although this did not appear to be an issue, it is unclear whether participants would have stepped elsewhere with their right leg. It is also possible that the biofeedback program monitoring lean magnitude, vertical forces and the center of pressure position may have introduced a cognitive load, creating a dual-task scenario. Oppositely, the biofeedback program may have prevented the participants from trying to anticipate the tether-release timing, as they were focused on the biofeedback program and meeting the pre-release requirements. The final limitation relates to only sampling young participants. Future work should build on this study to examine potential age- or pathology-related differences that might help to explain the high rates of falls in those populations. Although it is important to acknowledge these limitations, this was the first study to assess balance control and foot-placement during the landing phase of two-step responses.

In conclusion, this study investigated the landing phase during one and two-step responses, which is important as real-world perturbations such as tripping while walking often require multi-step responses. Differences were observed in COM\xCOM displacement between one and two-step conditions, as well as between the first and second step of two-step responses. Additionally, asymmetrical loading was found to influence both step length and width metrics. Furthermore, xCOM displacements between the first and second step were significantly correlated in the AP and ML directions. These results help to inform on the role of body movement after FC during reactive stepping, while the body weight symmetry results may have implications for individuals who exhibit body weight asymmetries during reactive stepping tasks.

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Next steps should focus on analyzing landing phase kinetics for single vs. two-step responses, and initial stepping limb dominance. Long-term goals should focus on studying landing phase control during more dynamic activities such as tripping while walking.

7. SUMMARY OF NOVELTY, CONTRIBUTION AND FUTURE DIRECTIONS

The four studies presented as part of this thesis are each novel in their own respects. Numerous studies have examined balance control during reactive stepping, with more recent work focusing on the landing phase occurring after foot-contact (FC) (Singer, et al., 2016). However, while these recent studies laid important ground work and provided important insights, many questions still remained about the control responses during this important phase of reactive stepping. Accordingly, while the results of the four studies presented in this thesis must be interpreted with caution due to the predictable nature of the tether-release paradigm (i.e., only forward-directed perturbations), and the instructional set employed, many novel insights were observed. The novelty, contribution and future directions of each study will be discussed below.

This first study included in this thesis (Chapter 3) was novel as it was the first to focus on whether center of mass (COM) displacement and velocity during the stepping phase can predict COM displacement during the landing phase (of reactive stepping). The first study also provided insight into whether individual characteristics can be used to predict COM displacement during the landing phase. The primary contributions from this study were the observations that COM displacement during the stepping and landing phases are correlated, and that specificity is important in predicting COM displacement during reactive stepping. Knowing that balance control is correlated between phases suggests that researchers and\or clinicians could focus on training the stepping phase, and as individuals improve their control, concomitant improvements may also be observed in the landing phase. Research has shown that a single session of perturbation-based training can result in improvements in stability at FC in community-dwelling older adults, which persist up to 12-months post training (Pai et al., 2014). A single session of perturbation-based training can also result in improvements in peak COM displacement during forward and backward reactive stepping, which persists for 24 hours (Dijkstra, et al., 2015). Second, this information could prove to be valuable for researchers and clinicians who are interested in training reactive stepping, as it suggests that measures calculated directly from reactive stepping trials are better

predictors of COM displacement, compared to independent variables calculated\measured separately from the reactive stepping task.

Regarding the general vs. specific predictors, overall the specific models resulted in stronger predictions of COM displacement in both age groups, at both points in time. The idea of training specificity is not new in the athletic world, where theoretically, the most specific and therefore functional form of training is to perform the actual movement(s) of the sport (Gamble, 2006; Siff, 2002). Perhaps the same is true in the domain of reactive balance training (Mansfield et al., 2010; Mansfield et al., 2015). While perturbation-based reactive stepping studies show promise for improving step metrics, and potentially reducing fall risk (Bhatt, et al., 2012; Mansfield, et al., 2010; Mansfield, et al., 2015; Pai, et al., 2014), these studies generally have not focused explicitly on landing phase control. As multiple studies suggest that age-related differences in initial spatio-temporal metrics are relatively small (McIlroy & Maki, 1996; Rogers, et al., 2001), future perturbation-based training studies should explore if this type of training has benefits for landing phase control. Lastly, as Study 1 found that hip extension strength was the only significant predictor of COM displacement at FC and the peak displacement after FC in the older adult general models, specifically targeting hip extension strength in older adults may also be a beneficial way to improve landing phase control in older adults. Recall, a study by Sibley et al. (2013) found that in Ontario physiotherapists, the three most commonly used clinical balance tools were the single leg stance, timed up and go (TUG), and Berg Balance Test. All three of these tests require components of hip extension strength, especially the TUG test. Therefore, focusing on hip extension strength (e.g., via squats) may not only facilitate landing phase control in older adults, but also improve other aspects of clinically-assessed balance control. Furthermore, beyond muscle strength, muscle power may be equally or more important in older adults for performing activities such as stair climbing, rising from a chair, and walking (Bassey et al., 1992; Evans, 2000). Recent studies of lateral balance recovery in older adults have also implicated the rate of force development as an important factor for distinguishing fallers and non-fallers, and the type of lateral stepping strategy used (Addison et al., 2017; Inacio et al., 2014). Accordingly, plyometric training in older adults incorporating hip extension movements (i.e.,

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countermovement vertical jumps, or drop landing vertical jumps) could be beneficial for improving

landing phase performance in older adults, as well as the execution of activities of daily living.

The thesis' second study (Chapter 4) was novel, as no published literature to date has reported

electromyographic (EMG) patterns during the landing phase of reactive stepping. This study compliments

Chapter 3, as researchers and clinicians who are interested in training the reactive stepping response

during a given phase could target the muscles which were most active in that specific phase. Accordingly,

the muscles which were active during the landing phase (compared to the earlier phases) were the biceps

femoris of the step-leg (which was correlated with the step-leg medial gastrocnemius peak magnitude

during the landing phase), and the rectus femoris and tibialis anterior of the support-leg. This is likely due

to the role of the step-leg biceps femoris as a hip extensor. In contrast, the rectus femoris of the support-

leg keeps the knee extended (and as a result help prevents the body from collapsing, likely in continuation

from the swing phase), while as discussed in section 4.5, the recruitment of the support-leg tibialis

anterior during phase 3 may be linked to the effectiveness of the biceps femoris and medial gastrocnemius

in resisting the initial rotational effects of the perturbation, and could serve to move the support-leg heel

back to the ground (if needed). While the mean peak timing and magnitude values which were analyzed

statistically, and presented in Tables 4-1 to 4-4 and Figure 4-4 provided novel and interesting insight into

muscle activation throughout the entirety of the reactive stepping response, the subject specific ensemble

averages (for each muscle) presented in Figure 4-5 suggest that future studies should focus on the

between-subject variability in muscle activation. Figure 4-5 shows that rectus femoris and tibialis anterior

activation was largely variable between participants. Specifically, both the step-leg tibialis anterior and

support-leg rectus femoris activation varied (visually) between-subjects over the entire time-frame

depicted in Figure 4-5. This may suggest that these specific muscles are particularly important for control

during the later aspects of the reactive stepping response (i.e., the landing phase). However, research is

needed to explore this idea further. Nonetheless, the results of this study contribute to the literature by

revealing which muscles were their most active during the landing phase, and which peak magnitudes

were correlated between muscles in a given phase. Focusing on the muscles (via exercise, etc.) which

were active during the landing phase could improve performance during this period, while understanding which muscle's peak magnitudes were correlated in a given phase could aid researchers in training recruitment patterns in older adults.

The thesis' third study (Chapter 5) provides new insights into whether wide stepping and arm movement are beneficial for balance control during the landing phase of reactive stepping. The results demonstrate that wide stepping is beneficial for stability in the medio-lateral (ML) direction, but may induce a larger anterior-posterior (AP) COM displacement, which could place greater demands on step length. Regarding arm movement, our results indicate that it may be more important for balance control in the ML direction, as opposed to the AP direction. In the ML direction, peak COM displacement after FC was larger in the restricted arm movement condition, when compared to the preferred stepping condition, regardless of age group. Accordingly, these findings support the potential value of incorporating arm movement training as part of re- or prehabilitation programs for those interested in preventing injuries due to sideways falls (i.e., hip fractures). While the current study did not focus on actual fall events, many researchers have previously suggested that future studies should focus on training protective arm reactions for fall-related injury prevention (Choi, et al., 2015; Feldman & Robinovitch, 2007; Schonnop, et al., 2013). Overall, the results will be important for helping to train balance recovery during forward reactive stepping, particularly in older adults, along with the results of Studies 1 and 2. However, with regards to the notion that wide stepping resulted in the largest ML stability margins in young and older adults, the benefits of wide stepping will have to be weighed with the fact that wide stepping is effortful, and humans tend to prefer a step width that minimizes metabolic cost (Donelan, et al., 2001). Anecdotally, many participants expressed that the wide stepping condition was the most physically taxing condition in Study 3.

The thesis' final study (Chapter 6) focused on the landing phase during two-step responses. A secondary objective was to quantify the effects of symmetrical and asymmetrical stance loading (pre-release). The novelty of Study 4 was that it was the first to focus on landing phase control during two-step responses, while also comparing two-step responses to one-step responses. When analyzing the first step,

two-step responses resulted in AP and ML COM displacements during the first step which were larger

and smaller, respectively, when compared to the one-step condition. With asymmetrical loading, first step

lengths were larger during the one-step condition (vs. two-step condition), while first step width was

reduced over both stepping tasks with asymmetrical loading. Peak AP extrapolated COM (xCOM)

displacement after foot-contact was larger in the second step, compared to the first step, with

asymmetrical loading. This suggests that asymmetrical loading may be beneficial when the goal is to

continue forward progression of the body, such as during gait. Interestingly, the first step resulted in a

greater ML xCOM displacement vs. the second step, regardless of loading, however, the first step width

was narrower (compared to the second step) with asymmetrical loading. Lastly, peak xCOM

displacements between the first and second steps were significantly correlated, and incorporating the first

step metrics from the one-step only condition did not strengthen these correlations. Overall, these results

suggest that single-step responses should not be used to infer balance control during multi-step scenarios.

This carries implications for those researchers interested in studying reactive stepping responses.

Although one-step responses may be more feasible to evoke safely, if one's goal is to gain insight into

balance control during multi-step scenarios, a perturbation large enough to evoke multiple steps should be

used. Or, alternatively, participants could be instructed to use two-steps, as recent research has shown that

instructing participants to recover with one or two reactive steps had little functional effect on the spatio-

temporal metrics of the first step (Cyr & Smeesters, 2009). However, as Study 4 revealed, there are

differences in landing phase balance control (COM\xCOM) when participants are instructed to respond

with one vs. two reactive steps.

While this thesis provides a complementary collection of studies that enhance our understanding

of balance control during reactive stepping responses, it also highlights future directions that could

provide further insights. Potential future directions would be to study the landing phase in a group of

older adults who are at a greater risk of falling than the community-dwelling older adults who participated

in the current study. Further novel insights may be gained about body movement after FC by studying

individuals such as frail older adults or individuals with neurological disorders such as Parkinson's

disease, or stroke patients. Additionally, as discussed in Chapter 3, studying a highly active sample of older adults, who are not 'fearful' or 'nervous' of the tether-release task could provide novel insights into the physiological effect of aging on reactive stepping, without the influence of psychological factors. One possibility to address the fact that the tether-release task can be daunting for some older adults, is to explore the potential of virtual reality (VR) in studying or training reactive stepping. The use of VR would eliminate the physical perturbation involved with the tether-release, surface translation, etc., which could be less physically demanding\fatiguing for older adults (compared to repeated exposure to an externally applied force). In turn, this would allow for an increased number of trials to be studied. To build on Chapter 3, a larger sample should be collected to assess the predictive utility of a larger number of predictor variables, while EMG recruitment patterns during the landing phase must be assessed in older adults to accompany the data presented here. This would inform researchers if there are obvious age-related discrepancies in lower-limb muscle recruitment which can be targeted clinically. Additionally, researchers should focus on studying landing phase balance control during more 'realistic' activities such as the rapid termination of gait and trip recovery during gait. Accounting for the effects of preceding gait will be important as a higher walking speed increases the effect of the perturbation, which can make recovery after tripping more difficult (Pijnappels, et al., 2008). Pavol et al. (2002) observed that a main contributor to falls during a reactive step was a faster walking speed, which paradoxically could place high strength older adults at an elevated risk of falling. Future studies should also assess whether the properties of the foot and\or toes (e.g., strength, flexibility, sensitivity, etc.) are important for landing phase control. Finally, researchers must aim to determine if landing phase control is related to actual fall-risk in older adults. To conclude, the novel contributions of the current thesis to the literature are new insights into the landing phase during reactive stepping, with a focus on predictive factors, muscle recruitment, movement restraints and two-step responses.

8. REFERENCES

Addison, O., Inacio, M., Bair, W. N., Beamer, B. A., Ryan, A. S., & Rogers, M. W. (2017). Role of hip abductor muscle composition and torque in protective stepping for lateral balance recovery in older adults. *Archives of Physical Medicine and Rehabilitation, 98*(6), 1223-1228.

Alexander, B. H., Rivara, F. P., & Wolf, M. E. (1992). The cost and frequency of hospitalization for fall-related injuries in older adults. *American Journal of Public Health, 82*(7), 1020-1023.

Allum, J. H. J., Carpenter, M. G., Honegger, F., Adkin, A. L., & Bloem, B. R. (2002). Age-dependent variations in the directional sensitivity of balance corrections and compensatory arm movements in man. *Journal of Physiology, 542*(2), 643-663.

Arampatzis, A., Karamanidis, K., & Mademli, L. (2008). Deficits in the way to achieve balance related to mechanisms of dynamic stability control in the elderly. *Journal of Biomechanics, 41*(8), 1754-1761.

Arampatzis, A., Peper, A., & Bierbaum, S. (2011). Exercise of mechanisms for dynamic stability control increases stability performance in the elderly. *Journal of Biomechanics, 44*(1), 52-58.

Barrett, R. S., Cronin, N. J., Lichtwark, G. A., Mills, P. M., & Carty, C. P. (2012). Adaptive recovery responses to repeated forward loss of balance in older adults. *Journal of Biomechanics, 45*(1), 183-187.

Bassey, E. J., Fiatarone, M. A., O'Neill, E. F., Kelly, M., Evans, W. J., & Lipsitz, L. A. (1992). Leg extensor power and functional performance in very old men and women. [Research Support, U.S. Gov't, Non-P.H.S.
Research Support, U.S. Gov't, P.H.S.]. *Clinical Science, 82*(3), 321-327.

Bhatt, T., Yang, F., & Pai, Y. C. (2012). Learning to resist gait-slip falls: long-term retention in community-dwelling older adults. *Archives of Physical Medicine and Rehabilitation, 93*(4), 557-564.

Brenneman, E. C. (2014). *Evaluation of biomechanical and neuromuscular effects of prophylactic knee brace use following exercise.* Master of Science, University of Waterloo, Waterloo, Ontario.

Brunt, D., Lafferty, M. J., McKeon, A., Goode, B., Mulhausen, C., & Polk, P. (1991). Invariant characteristics of gait initiation. *American Journal of Physical Medicine & Rehabilitation, 70*(4), 206-212.

Canada, S. (2010). Population Projections for Canada, Provinces and Territories: 2009 to 2036. Ottawa, Ontario: Statistics Canada.

Carbonneau, E., & Smeesters, C. (2014). Effects of age and lean direction on the threshold of single-step balance recovery in younger, middle-aged and older adults. *Gait and Posture, 39*(1), 365-371.

Carpenter, M. G., Frank, J. S., Winter, D. A., & Peysar, G. W. (2001). Sampling duration effects on centre of pressure summary measures. *Gait and Posture, 13*(1), 35-40.

Carpenter, M. G., Murnaghan, C. D., & Inglis, J. T. (2010). Shifting the balance: evidence of an exploratory role for postural sway. *Neuroscience, 171*(1), 196-204.

Carty, C. P., Barrett, R. S., Cronin, N. J., Lichtwark, G. A., & Mills, P. M. (2012a). Lower limb muscle weakness predicts use of a multiple- versus single-step strategy to recover from forward loss of balance in older adults. *The Journals of Gerontology. Series A, Biological Sciences and Medical Sciences, 67*(11), 1246-1252.

Carty, C. P., Cronin, N. J., Lichtwark, G. A., Mills, P. M., & Barrett, R. S. (2012b). Lower limb muscle moments and power during recovery from forward loss of balance in male and female single and multiple steppers. *Clinical Biomechanics, 27*(10), 1031-1037.

Carty, C. P., Cronin, N. J., Lichtwark, G. A., Mills, P. M., & Barrett, R. S. (2012c). Mechanisms of adaptation from a multiple to a single step recovery strategy following repeated exposure to forward loss of balance in older adults. *PLoS One, 7*(3), e33591.

Carty, C. P., Cronin, N. J., Nicholson, D., Lichtwark, G. A., Mills, P. M., Kerr, G., . . . Barrett, R. S. (2014). Reactive stepping behaviour in response to forward loss of balance predicts future falls in community-dwelling older adults. *Age and Ageing, 44*(1), 109-115.

Carty, C. P., Mills, P., & Barrett, R. S. (2011). Recovery from forward loss of balance in young and older adults using the stepping strategy. *Gait and Posture, 33*(2), 261-267.

Choi, W. J., Wakeling, J. M., & Robinovitch, S. N. (2015). Kinematic analysis of video-captured falls experienced by older adults in long-term care. *Journal of Biomechanics, 48*(6), 911-920.

Coronado, V. G., Thomas, K. E., Sattin, R. W., & Johnson, R. L. (2005). The CDC traumatic brain injury surveillance system: characteristics of persons aged 65 years and older hospitalized with a TBI. *Journal of Head Trauma Rehabilitation, 20*(3), 215-228.

Crenshaw, J. R., & Grabiner, M. D. (2014). The influence of age on the thresholds of compensatory stepping and dynamic stability maintenance. *Gait and Posture, 40*(3), 363-368.

Crenshaw, J. R., Rosenblatt, N. J., Hurt, C. P., & Grabiner, M. D. (2012). The discriminant capabilities of stability measures, trunk kinematics, and step kinematics in classifying successful and failed compensatory stepping responses by young adults. *Journal of Biomechanics 45*(1), 129-133.

Cronin, N. J., Barrett, R. S., Lichtwark, G., Mills, P. M., & Carty, C. P. (2013). Decreased lower limb muscle recruitment contributes to the inability of older adults to recover with a single step following a forward loss of balance. *Journal of Electromyography and Kinesiology, 23*(5), 1139-1144.

Cyr, M., & Smeesters, C. (2009). Kinematics of the threshold of balance recovery are not affected by instructions limiting the number of steps in younger adults. *Gait and Posture, 29*(4), 628-633.

de Leva, P. (1996). Adjustments to Zatsiorsky-Seluyanov's segment inertia parameters. *Journal of Biomechanics, 29*(9), 1223-1230.

Dean, J. C., Alexander, N. B., & Kuo, A. D. (2007). The effect of lateral stabilization on walking in young and old adults. *IEEE Transactions on Bio-Medical Engineering, 54*(11), 1919-1926.

Dijkstra, B. W., Horak, F. B., Kamsma, Y. P., & Peterson, D. S. (2015). Older adults can improve compensatory stepping with repeated postural perturbations. *Frontiers in Aging Neuroscience, 7*, 201.

Do, M. C., Breniere, Y., & Brenguier, P. (1982). A biomechanical study of balance recovery during the fall forward. *Journal of Biomechanics, 15*(12), 933-939.

Donelan, J. M., Kram, R., & Kuo, A. D. (2001). Mechanical and metabolic determinants of the preferred step width in human walking. *Proceedings of the Royal Society of London B: Biological Sciences, 268*(1480), 1985-1992.

Donelan, J. M., Kram, R., & Kuo, A. D. (2002). Mechanical work for step-to-step transitions is a major determinant of the metabolic cost of human walking. *The Journal of Experimental Biology, 205*(Pt 23), 3717-3727.

Evans, J. D. (1996). *Straightforward statistics for the behavioral sciences.* Pacific Grove, CA: Brooks/Cole Publishing.

Evans, W. J. (2000). Exercise strategies should be designed to increase muscle power. [Comment Editorial]. *The Journals of Gerontology. Series A, Biological Sciences and Medical Sciences, 55*(6), M309-M310.

Feldman, F., & Robinovitch, S. N. (2007). Reducing hip fracture risk during sideways falls: evidence in young adults of the protective effects of impact to the hands and stepping. *Journal of Biomechanics, 40*(12), 2612-2618.

Fujimoto, M., Bair, W. N., & Rogers, M. W. (2017). Single and multiple step balance recovery responses can be different at first step lift-off following lateral waist-pull perturbations in older adults. *Journal of Biomechanics 55*, 41-47.

Gamble, P. (2006). Implications and applications of training specificity for coaches and athletes. *Strength and Conditioning Journal, 28*(3), 54-58.

Gilasi, H. R., Soori, H., Yazdani, S., & Taheri Tenjani, P. (2015). Fall-related injuries in community-dwelling older adults in Qom Province, Iran, 2010-2012. *Archives of Trauma Research, 4*(1), e22925.

Glinka, M. N. (2013). *The effect of aging on movement characteristics and postural control during stooping and crouching tasks.* Master of Science, University of Waterloo, Waterloo, Ontario.

Grabiner, M. D., Koh, T. J., Lundin, T. M., & Jahnigen, D. W. (1993). Kinematics of recovery from a stumble. *Journal of Gerontology, 48*(3), M97-102.

Grabiner, M. D., Owings, T. M., & Pavol, M. J. (2005). Lower extremity strength plays only a small role in determining the maximum recoverable lean angle in older adults. *The Journals of Gerontology. Series A, Biological Sciences and Medical Sciences, 60*(11), 1447-1450.

Grabiner, P. C., Biswas, S. T., & Grabiner, M. D. (2001). Age-related changes in spatial and temporal gait variables. *Archives of Physical Medicine and Rehabilitation, 82*(1), 31-35.

Graham, D. F., Carty, C. P., Lloyd, D. G., & Barrett, R. S. (2015). Biomechanical predictors of maximal balance recovery performance amongst community-dwelling older adults. *Experimental Gerontology, 66*, 39-46. doi: 10.1016/j.exger.2015.04.006

Graham, D. F., Carty, C. P., Lloyd, D. G., Lichtwark, G. A., & Barrett, R. S. (2014). Muscle contributions to recovery from forward loss of balance by stepping. *Journal of Biomechanics, 47*(3), 667-674.

Greenspan, S. L., Myers, E. R., Kiel, D. P., Parker, R. A., Hayes, W. C., & Resnick, N. M. (1998). Fall direction, bone mineral density, and function: risk factors for hip fracture in frail nursing home elderly. *The American Journal of Medicine, 104*(6), 539-545.

Harper, N. G., Esposito, E. R., Wilken, J. M., & Neptune, R. R. (2014). The influence of ankle-foot orthosis stiffness on walking performance in individuals with lower-limb impairments. *Clinical Biomechanics, 29*(8), 877-884.

Hase, K., & Stein, R. B. (1998). Analysis of rapid stopping during human walking. *Journal of Neurophysiology, 80*(1), 255-261.

Hof, A. L., Gazendam, M. G. J., & Sinke, W. E. (2005). The condition for dynamic stability. *Journal of Biomechanics, 38*(1), 1-8.

Holliday, P. J., Fernie, G. R., Gryfe, C. I., & Griggs, G. T. (1990). Video recording of spontaneous falls of the elderly. In B. E. Gray (Ed.), *Slips, Stumbles, and Falls: Pedestrian Footwear and Surfaces (ASTM STP 1103)* (pp. 7-16). Philadelphia: American Society for Testing and Materials.

Hsiao-Wecksler, E. T. (2008). Biomechanical and age-related differences in balance recovery using the tether-release method. *Journal of Electromyography and Kinesiology, 18*(2), 179-187.

Hsiao, E. T., & Robinovitch, S. N. (1998). Common protective movements govern unexpected falls from standing height. *Journal of Biomechanics, 31*(1), 1-9.

Hsiao, E. T., & Robinovitch, S. N. (1999). Biomechanical influences on balance recovery by stepping. *Journal of Biomechanics 32*(10), 1099-1106.

Hurt, C. P., & Grabiner, M. D. (2015). Age-related differences in the maintenance of frontal plane dynamic stability while stepping to targets. *Journal of Biomechanics 48*(4), 592-597.

Inacio, M., Ryan, A. S., Bair, W. N., Prettyman, M., Beamer, B. A., & Rogers, M. W. (2014). Gluteal muscle composition differentiates fallers from non-fallers in community dwelling older adults. [Comparative Study
Research Support, N.I.H., Extramural
Research Support, Non-U.S. Gov't]. *BMC Geriatrics, 14*, 37.

Johnson, S., Kelly, S., & Rasali, D. (2015). Differences in fall injury hospitalization and related survival rates among older adults across age, sex, and areas of residence in Canada. *Injury Epidemiology, 2*(1), 24.

Karamanidis, K., & Arampatzis, A. (2007). Age-related degeneration in leg-extensor muscle–tendon units decreases recovery performance after a forward fall: compensation with running experience. *European Journal of Applied Physiology, 99*(1), 73-85.

Karamanidis, K., Arampatzis, A., & Mademli, L. (2008). Age-related deficit in dynamic stability control after forward falls is affected by muscle strength and tendon stiffness. *Journal of Electromyography and Kinesiology, 18*(6), 980-989.

Kaya, B. K., Krebs, D. E., & Riley, P. O. (1998). Dynamic stability in elders: momentum control in locomotor ADL. *The Journals of Gerontology. Series A, Biological Sciences and Medical Sciences, 53*(2), M126-134.

King, G. W., Akula, C. K., & Luchies, C. W. (2012). Age-related differences in kinetic measures of landing phase lateral stability during a balance-restoring forward step. *Gait and Posture, 35*(3), 440-445.

Kurz, I., Berezowski, E., & Melzer, I. (2013). Frontal plane instability following rapid voluntary stepping: effects of age and a concurrent cognitive task. *The Journals of Gerontology. Series A, Biological Sciences and Medical Sciences, 68*(11), 1402-1408.

Lakhani, B., Mansfield, A., Inness, E. L., & McIlroy, W. E. (2011). Characterizing the determinants of limb preference for compensatory stepping in healthy young adults. *Gait and Posture, 33*(2), 200-204.

Lee, P. Y., Gadareh, K., & Bronstein, A. M. (2014). Forward–backward postural protective stepping responses in young and elderly adults. *Human Movement Science, 34*, 137-146.

Lockhart, T. E., & Kim, S. (2006). Relationship between hamstring activation rate and heel contact velocity: factors influencing age-related slip-induced falls. *Gait and Posture, 24*(1), 23-34.

Luchies, C. W., Alexander, N. B., Schultz, A. B., & Ashton-Miller, J. (1994). Stepping responses of young and old adults to postural disturbances: kinematics. *Journal of the American Geriatrics Society, 42*(5), 506-512.

Madigan, M. L., & Lloyd, E. M. (2005a). Age-related differences in peak joint torques during the support phase of single-step recovery from a forward fall. *Journal of Gerontology. Series A, Biological sciences and Medical Sciences, 60*(7), 910-914.

Madigan, M. L., & Lloyd, E. M. (2005b). Age and stepping limb performance differences during a single-step recovery from a forward fall. *The Journals of Gerontology. Series A, Biological Sciences and Medical Sciences, 60*(4), 481-485.

Maki, B. E. (1997). Gait changes in older adults: predictors of falls or indicators of fear. *Journal of American Geriatrics Society, 45*(3), 313-320.

Maki, B. E., Edmondstone, M. A., & McIlroy, W. E. (2000). Age-related differences in laterally directed compensatory stepping behavior. *The Journals of Gerontology. Series A, Biological Sciences and Medical Sciences, 55*(5), M270-M277.

Maki, B. E., Edmondstone, M. A., Perry, S. D., Heung, E., & Quant, S. (2001). Control of rapid limb movements for balance recovery: do age-related changes predict falling risk? In J. Duysens, B. C. M. Smits-Engelsman & H. Kingma (Eds.), *Control of Posture and Gait* (pp. 126-129). Maastricht, Netherlands: International Society for Postural and Gait Research.

Maki, B. E., & McIlroy, W. E. (1997). The role of limb movements in maintaining upright stance: the "change-in-support" strategy. *Physical Therapy, 77*(5), 488-507.

Maki, B. E., & McIlroy, W. E. (2006). Control of rapid limb movements for balance recovery: age-related changes and implications for fall prevention. *Age and Ageing, 35*(S2), 12-18.

Mansfield, A., Inness, E. L., Lakhani, B., & McIlroy, W. E. (2012). Determinants of limb preference for initiating compensatory stepping poststroke. *Archives of Physical Medicine and Rehabilitation, 93*(7), 1179-1184.

Mansfield, A., & Maki, B. E. (2009). Are age-related impairments in change-in-support balance reactions dependent on the method of balance perturbation? *Journal of Biomechanics, 42*(8), 1023-1031.

Mansfield, A., Peters, A. L., Liu, B. A., & Maki, B. E. (2010). Effect of a perturbation-based balance training program on compensatory stepping and grasping reactions in older adults: a randomized controlled trial. *Physical Therapy, 90*(4), 476-491.

Mansfield, A., Wong, J. S., Bryce, J., Knorr, S., & Patterson, K. K. (2015). Does perturbation-based balance training prevent falls? Systematic review and meta-analysis of preliminary randomized controlled trials. *Physical Therapy, 95*(5), 700-709.

McIlroy, W. E., & Maki, B. E. (1993a). Changes in early 'automatic' postural responses associated with the prior-planning and execution of a compensatory step. *Brain Research, 631*(2), 203-211.

McIlroy, W. E., & Maki, B. E. (1993b). Task constraints on foot movement and the incidence of compensatory stepping following perturbation of upright stance. *Brain Research, 616*(1-2), 30-38.

McIlroy, W. E., & Maki, B. E. (1995). Adaptive changes to compensatory stepping responses. *Gait and Posture, 3*(1), 43-50.

McIlroy, W. E., & Maki, B. E. (1996). Age-related changes in compensatory stepping in response to unpredictable perturbations. *The Journals of Gerontology. Series A, Biological Sciences and Medical Sciences, 51*(6), M289-296.

McIlroy, W. E., & Maki, B. E. (1997). Preferred placement of the feet during quiet stance: development of a standardized foot placement for balance testing. *Clinical Biomechanics (Bristol, Avon), 12*(1), 66-70.

McIlroy, W. E., & Maki, B. E. (1999). The control of lateral stability during rapid stepping reactions evoked by antero-posterior perturbation: does anticipatory control play a role? [Clinical Trial Randomized Controlled Trial Research Support, Non-U.S. Gov't]. *Gait and Posture, 9*(3), 190-198.

Menant, J. C., Steele, J. R., Menz, H. B., Munro, B. J., & Lord, S. R. (2009). Rapid gait termination: effects of age, walking surfaces and footwear characteristics. *Gait and Posture, 30*(1), 65-70.

Mille, M. L., Johnson-Hilliard, M., Martinez, K. M., Zhang, Y., Edwards, B. J., & Rogers, M. W. (2013). One step, two steps, three steps more ... directional vulnerability to falls in community-dwelling older people. *The Journals of Gerontology. Series A, Biological Sciences and Medical Sciences, 68*(12), 1540-1548.

Milner-Brown, H. S., Stein, R. B., & Yemm, R. (1973a). The contractile properties of human motor units during voluntary isometric contractions. *Journal of Physiology, 228*(2), 285-306.

Milner-Brown, H. S., Stein, R. B., & Yemm, R. (1973b). The orderly recruitment of human motor units during voluntary isometric contractions. *Journal of Physiology, 230*(2), 359-370.

Murnaghan, C. D., Squair, J. W., Chua, R., Inglis, J. T., & Carpenter, M. G. (2013). Are increases in COP variability observed when participants are provided explicit verbal cues prior to COM stabilization? *Gait and Posture, 38*(4), 734-738.

Nevitt, M. C., & Cummings, S. R. (1993). Type of fall and risk of hip and wrist fractures: the study of osteoporotic fractures. The Study of Osteoporotic Fractures Research Group. *Journal of the American Geriatric Society, 41*(11), 1226-1234.

Nevitt, M. C., Cummings, S. R., Kidd, S., & Black, D. (1989). Risk factors for recurrent nonsyncopal falls. A prospective study. *The Journal of the American Medical Association, 261*(18), 2663-2668.

Nüesch, C., Laffer, D., Netzer, C., Pagenstert, G., & Mündermann, A. (2016). Effect of gait retraining for reducing ambulatory knee load on trunk biomechanics and trunk muscle activity. *Gait and Posture, 47*, 24-30.

Nussbaum, M. A., & Zhang, X. (2000). Heuristics for locating upper extremity joint centres from a reduced set of surface markers. *Human Movement Science, 19*(5), 797-816.

O'Loughlin, J. L., Robitaille, Y., Boivin, J. F., & Suissa, S. (1993). Incidence of and risk factors for falls and injurious falls among the community-dwelling elderly. *American Journal of Epidemiology, 137*(3), 342-354.

O'Connell, C., Chambers, A., Mahboobin, A., & Cham, R. (2016). Effects of slip severity on muscle activation of the trailing leg during an unexpected slip. *Journal of Electromyography and Kinesiology, 28*, 61-66.

O'Connor, C. M., Thorpe, S. K., O'Malley, M. J., & Vaughan, C. L. (2007). Automatic detection of gait events using kinematic data. *Gait and Posture, 25*(3), 469-474.

Ochi, A., Yokoyama, S., Abe, T., Yamada, K., Tateuchi, H., & Ichihashi, N. (2014). Differences in muscle activation patterns during step recovery in elderly women with and without a history of falls. *Aging Clinical and Experimental Research, 26*(2), 213-220.

Pai, Y. C., Maki, B. E., Iqbal, K., McIlroy, W. E., & Perry, S. D. (2000). Thresholds for step initiation induced by support-surface translation: a dynamic center-of-mass model provides much better prediction than a static model. *Journal of Biomechanics, 33*(3), 387-392.

Pai, Y. C., & Patton, J. (1997). Center of mass velocity-position predictions for balance control. *Journal of Biomechanics, 30*(4), 347-354.

Pai, Y. C., Rogers, M. W., Patton, J., Cain, T. D., & Hanke, T. A. (1998). Static versus dynamic predictions of protective stepping following waist-pull perturbations in young and older adults. *Journal of Biomechanics, 31*(12), 1111-1118.

Pai, Y. C., Yang, F., Bhatt, T., & Wang, E. (2014). Learning from laboratory-induced falling: long-term motor retention among older adults. *Age, 36*(3), 1367-1376.

Parijat, P., Lockhart, T. E., & Liu, J. (2015). EMG and kinematic responses to unexpected slips after slip training in virtual reality. *IEEE Transactions on Biomedical Engineering, 62*(2), 593-599.

Patterson, K. K., Parafianowicz, I., Danells, C. J., Closson, V., Verrier, M. C., Staines, W. R., . . . McIlroy, W. E. (2008). Gait asymmetry in community-ambulating stroke survivors. *Archives of Physical Medicine and Rehabilitation, 89*(2), 304-310.

Pavol, M. J., Owings, T. M., Foley, K. T., & Grabiner, M. D. (2001). Mechanisms leading to a fall from an induced trip in healthy older adults. *The Journals of Gerontology. Series A, Biological Sciences and Medical Sciences, 56*(7), M428-437.

Pavol, M. J., Owings, T. M., Foley, K. T., & Grabiner, M. D. (2002). Influence of lower extremity strength of healthy older adults on the outcome of an induced trip. *Journal of the American Geriatrics Society, 50*(2), 256-262.

PHAC. (2005). Report on Seniors' Falls in Canada. Ottawa, Ontario: Public Health Agency of Canada, Division of Aging and Seniors.

Pickett, W., Ardern, C., & Brison, R. J. (2001). A population-based study of potential brain injuries requring emergency care. *Canadian Medical Association Journal, 165*(3), 288-292.

Pijnappels, M., Bobbert, M. F., & van Dieën, J. H. (2004). Contribution of the support limb in control of angular momentum after tripping. *Journal of Biomechanics, 37*(12), 1811-1818.

Pijnappels, M., Bobbert, M. F., & van Dieën, J. H. (2005). How early reactions in the support limb contribute to balance recovery after tripping. *Journal of Biomechanics, 38*(3), 627-634.

Pijnappels, M., Reeves, N. D., Maganaris, C. N., & van Dieën, J. H. (2008). Tripping without falling; lower limb strength, a limitation for balance recovery and a target for training in the elderly. *Journal of Electromyography and Kinesiology, 18*(2), 188-196.

Porter, S., & Nantel, J. (2015). Older adults prioritize postural stability in the anterior-posterior direction to regain balance following volitional lateral step. *Gait and Posture, 41*(2), 666-669. doi: 10.1016/j.gaitpost.2015.01.021
S0966-6362(15)00024-7 [pii]

Raina, P., Dukeshire, S., Chambers, L., Toivonen, D., & Lindsay, J. (1997). Prevalence, risk factors, and health care utilization for injuries among Canadian seniors: An analysis of 1994 National Population Health Survey (IESOP Research Report No. 15). . Hamilton, Ontario: McMaster University.

Robinovitch, S. N., Feldman, F., Yang, Y., Schonnop, R., Leung, P. M., Sarraf, T., . . . Loughin, M. (2013). Video capture of the circumstances of falls in elderly people residing in long-term care: an observational study. *Lancet, 381*(9860), 47-54.

Robinovitch, S. N., Normandin, S. C., Stotz, P., & Maurer, J. D. (2005). Time requirement for young and elderly women to move into a position for breaking a fall with outstretched hands. *The Journals of Gerontology. Series A, Biological Sciences and Medical Sciences, 60*(12), 1553-1557.

Rogers, M. W., Hedman, L. D., Johnson, M. E., Cain, T. D., & Hanke, T. A. (2001). Lateral stability during forward-induced stepping for dynamic balance recovery in young and older adults. *The Journals of Gerontology. Series A, Biological Sciences and Medical Sciences, 56*(9), M589-594.

Schonnop, R., Yang, Y., Feldman, F., Robinson, E., Loughin, M., & Robinovitch, S. N. (2013). Prevalence of and factors associated with head impact during falls in older adults in long-term care. *Canadian Medical Association Journal, 185*(17), E803-810.

Schrager, M. A., Kelly, V. E., Price, R., Ferrucci, L., & Shumway-Cook, A. (2008). The effects of age on medio-lateral stability during normal and narrow base walking. *Gait and Posture, 28*(3), 466-471.

Schulz, B. W., Ashton-Miller, J. A., & Alexander, N. B. (2005). Compensatory stepping in response to waist pulls in balance-impaired and unimpaired women. *Gait and Posture, 22*(3), 198-209.

Schwartz, A. V., Kelsey, J. L., Sidney, S., & Grisso, J. A. (1998). Characteristics of falls and risk of hip fracture in elderly men. *Osteoporosis International, 8*(3), 240-246.

Serrao, M., Conte, C., Casali, C., Ranavolo, A., Mari, S., Di Fabio, R., . . . Pierelli, F. (2013). Sudden stopping in patients with cerebellar ataxia. *Cerebellum, 12*(5), 607-616.

Sibley, K. M., Straus, S. E., Inness, E. L., Salbach, N. M., & Jaglal, S. B. (2013). Clinical balance assessment: perceptions of commonly-used standardized measures and current practices among physiotherapists in Ontario, Canada. [Research Support, Non-U.S. Gov't]. *Implementation Science, 8*, 33. doi: 10.1186/1748-5908-8-33

Siff, M. C. (2002). Functional training revisited. *Strength and Conditioning Journal, 24*(5), 42-46.

Singer, J. C. (2012). *Age-related changes in the control of mediolateral dynamic stability during volitional and reactive stepping.* Doctor of Philosophy, University of Waterloo, Waterloo, Ontario, Canada.

Singer, J. C., McIlroy, W. E., & Prentice, S. D. (2014). Kinetic measures of restabilisation during volitional stepping reveal age-related alterations in the control of mediolateral dynamic stability. *Journal of Biomechanics, 47*(14), 3539-3545.

Singer, J. C., Prentice, S. D., & McIlroy, W. E. (2012). Dynamic stability control during volitional stepping: a focus on the restabilisation phase at movement termination. *Gait and Posture, 35*(1), 106-110.

Singer, J. C., Prentice, S. D., & McIlroy, W. E. (2013). Age-related changes in mediolateral dynamic stability control during volitional stepping. *Gait and Posture, 38*(4), 679-683.

Singer, J. C., Prentice, S. D., & McIlroy, W. E. (2016). Age-related challenges in reactive control of mediolateral stability during compensatory stepping: a focus on the dynamics of restabilisation. *Journal of Biomechanics, 49*(5), 749-755.

SMARTRISK. (2009). The Economic Burden of Injury in Canada. Toronto, Ontario: SMARTRISK.

Sparrow, W. A., & Tirosh, O. (2003). Identifying heel contact and toe-off using forceplate thresholds with a range of digital-filter cutoff frequencies. *Journal of Applied Biomechanics, 19*(2), 178-184.

Tajeu, G. S., Delzell, E., Smith, W., Arora, T., Curtis, J. R., Saag, K. G., . . . Kilgore, M. L. (2014). Death, debility, and destitution following hip fracture. *The Journals of Gerontology. Series A, Biological Sciences and Medical Sciences, 69*(3), 346-353.

Thelen, D. G., Wojcik, L. A., Schultz, A. B., Ashton-Miller, J. A., & Alexander, N. B. (1997). Age differences in using a rapid step to regain balance during a forward fall. *The Journals of Gerontology. Series A, Biological Sciences and Medical Sciences, 52*(1), M8-13.

Thomas, K. E., Stevens, J. A., Sarmiento, K., & Wald, M. M. (2008). Fall-related traumatic brain injury deaths and hospitalizations among older adults — United States, 2005. *Journal of Safety Research, 39*(3), 269-272.

Tinetti, M. E., Mendes de Leon, C. F., Doucette, J. T., & Baker, D. I. (1994). Fear of falling and fall-related efficacy in relationship to functioning among community-living elders. *Journal of Gerontology, 49*(3), M140-147.

Tinetti, M. E., Speechley, M., & Ginter, S. F. (1988). Risk factors for falls among elderly persons living in the community. *The New England Journal of Medicine, 319*(26), 1701-1707.

Tirosh, O., & Sparrow, W. A. (2004). Gait termination in young and older adults: effects of stopping stimulus probability and stimulus delay. *Gait and Posture, 19*(3), 243-251.

Tirosh, O., & Sparrow, W. A. (2005). Age and walking speed effects on muscle recruitment in gait termination. *Gait and Posture, 21*(3), 279-288.

Tokuno, C. D., Cresswell, A. G., Thorstensson, A., & Carpenter, M. G. (2010). Age-related changes in postural responses revealed by support-surface translations with a long acceleration–deceleration interval. *Clinical Neurophysiology, 121*(1), 109-117.

Tromp, A. M., Pluijm, S. M., Smit, J. H., Deeg, D. J., Bouter, L. M., & Lips, P. (2001). Fall-risk screening test: a prospective study on predictors for falls in community-dwelling elderly. *Journal of Clinical Epidemiology, 54*(8), 837-844.

Troy, K. L., Donovan, S. J., Marone, J. R., Bareither, M. L., & Grabiner, M. D. (2008). Modifiable performance domain risk-factors associated with slip-related falls. *Gait and Posture, 28*(3), 461-465.

Vallee, P., Tisserand, R., & Robert, T. (2015). Possible recovery or unavoidable fall? A model to predict the one step balance recovery threshold and its stepping characteristics. *Journal of Biomechanics 48*(14), 3905-3911.

Viggiani, D. (2015). *The effects of hip abductor fatigue on low back pain development during prolonged standing.* Master of Science, University of Waterloo, Waterloo, Ontario.

Weaver, T. B., Hamilton, L. E., & Tokuno, C. D. (2012). Age-related changes in the control of perturbation-evoked and voluntary arm movements. *Clinical Neurophysiology, 123*(10), 2025-2033.

Weerdesteyn, V., Laing, A. C., & Robinovitch, S. N. (2008). Automated postural responses are modified in a functional manner by instruction. *Experimental Brain Research, 186*(4), 571-580.

Weinhandl, J. T., & O'Connor, K. M. (2010). Assessment of a greater trochanter-based method of locating the hip joint center. *Journal of Biomechanics, 43*(13), 2633-2636.

WHO. (2007). Global Report on Falls Prevention in Older Age. Geneva, Switzerland: World Health Organization.

Winter, D. A. (1995). Human balance and posture control during standing and walking. *Gait and Posture, 3*(1), 193-214.

Winter, D. A. (2009). *Biomechanics and Motor Control of Human Movement.* Hoboken, New Jersey: John Wiley & Sons, Inc.

Winter, D. A., Patla, A. E., Frank, J. S., & Walt, S. E. (1990). Biomechanical walking pattern changes in the fit and healthy elderly. *Physical Therapy, 70*(6), 340-347.

Winter, D. A., & Yack, H. J. (1987). EMG profiles during normal human walking: stride-to-stride and inter-subject variability. *Electroencephalography and Clinical Neurophysiology, 67*(5), 402-411.

Wojcik, L. A., Thelen, D. G., Schultz, A. B., Ashton-Miller, J. A., & Alexander, N. B. (1999). Age and gender differences in single-step recovery from a forward fall. *The Journals of Gerontology. Series A, Biological Sciences and Medical Sciences, 54*(1), M44-50.

Wright, A. D., Heckman, G. A., McIlroy, W. E., & Laing, A. C. (2014). Novel safety floors do not influence early compensatory balance reactions in older adults. *Gait and Posture, 40*(1), 160-165.

Wu, G., & Cavanagh, P. R. (1995). ISB recommendations for standardization in the reporting of kinematic data. *Journal of Biomechanics, 28*(10), 1257-1261.

Yang, J. F., & Winter, D. A. (1984). Electromyographic amplitude normalization methods: improving their sensitivity as diagnostic tools in gait analysis. *Archives of Physical Medicine and Rehabilitation, 65*(9), 517-521.

Yang, Y., Mackey, D. C., Liu-Ambrose, T., Feldman, F., & Robinovitch, S. N. (2015). Risk factors for hip impact during real-life falls captured on video in long-term care. *Osteoporosis International*.

Yang, Y., Schonnop, R., Feldman, F., & Robinovitch, S. N. (2013). Development and validation of a questionnaire for analyzing real-life falls in long-term care captured on video. *BMC Geriatrics, 13*(1), 40-50.

9. APPENDICES

9.1 *Appendix 1 – Telephone Interview Questions:*

All participants were interviewed prior to participation to ensure they:

1) were able to stand, without moving, for at least 60 seconds;

2) were confident in their ability to independently perform the balance recovery task involving the use of

a forward step;

3) were able to walk and stand without the use of a cane, walker or other ambulatory aid;

4) were free of any health issues/conditions which could affect their balance;

5) were not using any psychotropic medications;

6) were free of any current injuries to their torso, legs or arms;

7) did not have or experience a heart condition; pacemaker, balance issues, prone to dizziness, light

headedness, or fainting due to medications;

8) were free of any allergy or sensitivity to medical tape and running alcohol

Vertex of the head

External auditory meatus

Gonion process

Acromio-clavicular joint

Tubercle of the humerus

Epicondyles of the humerus

Iliac crest

Greater trochanter of the femur

Styloid processes of the wrist

3rd metacarpal

Epicondlyes of the femur

Condyles of the tibia

Malleoli of the tibia

Head of the 5th metatarsal

Head of the 2nd metatarsal

Tip of the great toe

Calcaneus
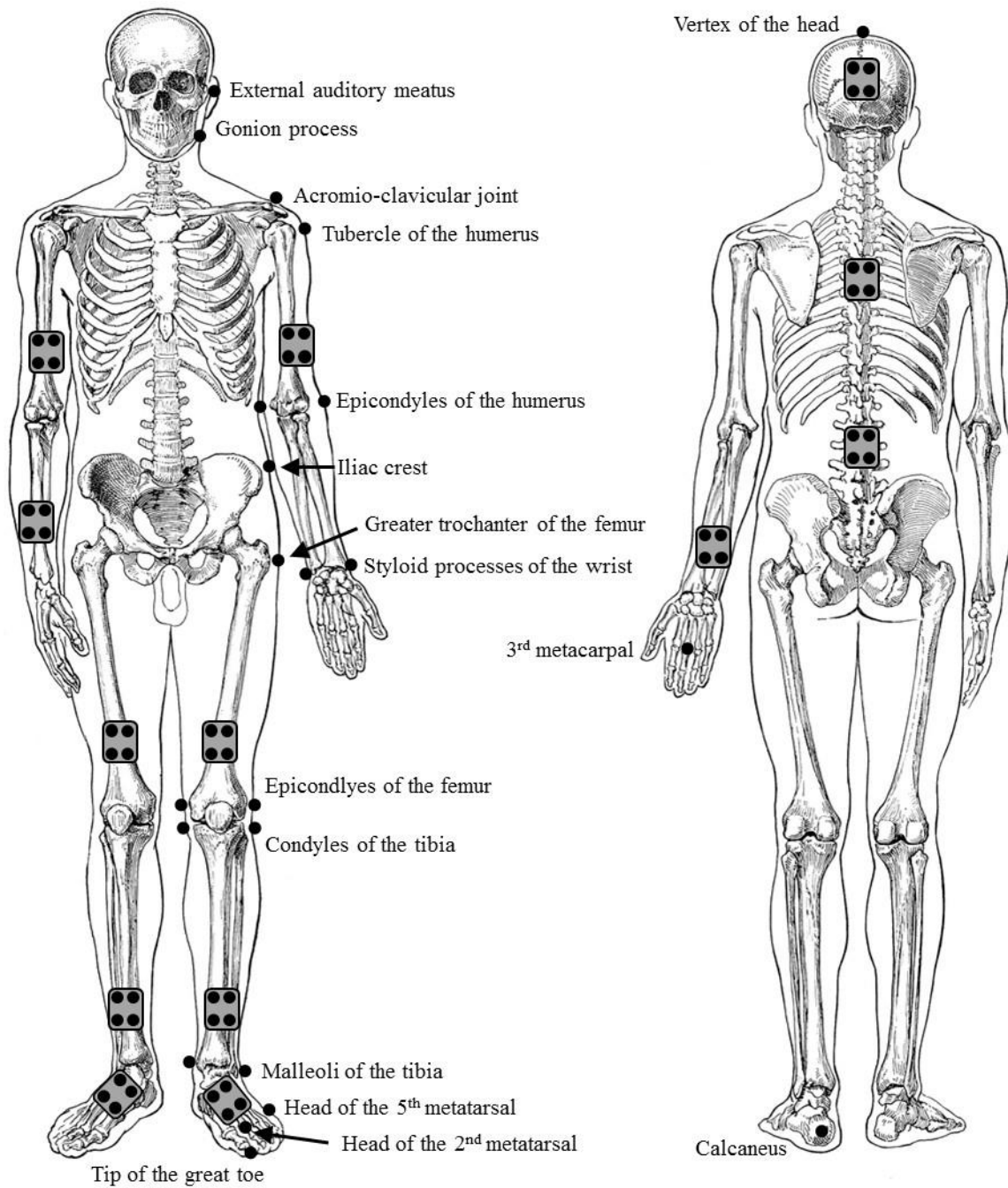
http://etc.usf.edu/clipart/52300/52330/52330_skeleton_lg.gif

9.3   *Appendix 3 – Supplementary Tables*

Supplementary Table 1: Mean(SD) values for the initial conditions for the first step comparison. Note that the stability margins reflect the period prior to cable-release. 2 x 2 repeated measures ANOVAs were used with stepping task (one-step vs. two-steps) and body weight symmetry (symmetrical vs. asymmetrical) to assess for statistically significant effects.

	One-step, Symmetrical BW%	One-step, Asymmetrical BW%	Two-steps, Symmetrical BW%	Two-steps, Asymmetrical BW%	Effect(s); *p*	Mean Difference
Tether Load (%)	12.8(0.8)	12.8(0.6)	12.8(0.6)	12.9(0.6)	Stepping Task; *p*=0.735 Body Weight; *p*=0.774 Interaction; *p*=0.507	---------------
Right Vert. Force (%)	49.9(0.6)	40.1(0.6)	50.4(0.6)	40.3(0.5)	Stepping Task; ***p*=0.007** Body Weight; ***p*<0.001** Interaction; *p*=0.070	0.29 % 9.95 %
Left Vert. Force (%)	50.1(0.6)	59.9(0.6)	49.6(0.6)	59.7(0.5)	Step Number; ***p*=0.007** Body Weight; ***p*<0.001** Interaction; *p*=0.070	0.29 % 9.95 %
AP Stability Margin (mm)	-44.8(22.4)	-43.6(21.8)	-46.4(22.3)	-45.7(20.9)	Stepping Task; *p*=0.300 Body Weight; *p*=0.587 Interaction; *p*=0.841	---------------
ML Stability Margin (mm)	209.1(11.2)	240.1(12.1)	206.1(11.1)	240.4(12.5)	Stepping Task; *p*=0.057 Body Weight; ***p*<0.001** Interaction; *p*=0.078	32.69 mm

BW = body weight; AP = anterior-posterior; ML = medio-lateral.

Supplementary Table 2: Mean(SD) values for the two-step response comparison. Paired-samples t-tests were used to compare if the initial conditions differed between the two loading conditions.

	Symmetrical BW%	Asymmetrical BW%	*p*-value
Tether Load (%)	12.8 (0.6)	12.9(0.6)	*p*=0.528
Right Vert. Force (%)	50.4(0.6)	40.3(0.5)	***p*<0.001**
Left Vert. Force (%)	49.6 (0.6)	59.7(0.5)	***p*<0.001**
AP Stability Margin (mm)	-46.4(22.3)	-45.7(20.9)	*p*=0.764
ML Stability Margin (mm)	206.1(11.1)	240.4(12.4)	***p*<0.001**

BW = body weight.

Supplementary Table 3: Mean(SD) values for the peak AP COM displacement after foot-contact during the first step. A 2 x 2 repeated measures ANOVA was used with stepping task (one-step vs. two-steps) and body weight symmetry (symmetrical vs. asymmetrical) as factors. In the case of a significant interaction, a paired-samples t-test was used for post-hoc analysis.

Step Condition	Body Weight Symmetry		Average (mm)
	Symmetrical	**Asymmetrical**	
One-Step (mm)	274.2(32.3)	288.3(28.2)	281.2(30.7)
Two-Steps (mm)	306.8(30.9)	308.3(29.8)	307.6(29.9)
Average (mm)	290.5(35.3)	298.3(30.3)	-----------------

AP = anterior-posterior; COM = center of mass.

Supplementary Table 4: Mean(SD) values for the peak ML COM displacement after foot-contact during the first step. A 2 x 2 repeated measures ANOVA was used with stepping task (one-step vs. two-steps) and body weight symmetry (symmetrical vs. asymmetrical) as factors. In the case of a significant interaction, a paired-samples t-test was used for post-hoc analysis.

| Step Condition | Body Weight Symmetry | | Average (mm) |
	Symmetrical	Asymmetrical	
One-Step (mm)	83.3(13.3)	90.7(15.8)	87.0(14.9)
Two-Steps (mm)	60.4(12.0)	56.5(15.7)	58.5(13.9)
Average (mm)	71.9(17.0)	73.6(23.3)	-----------------

ML = medio-lateral; COM = center of mass.

Supplementary Table 5: Mean(SD) values for step length (of the first step). A 2 x 2 repeated measures ANOVA was used with stepping task (one-step vs. two-steps) and body weight symmetry (symmetrical vs. asymmetrical) as factors. In the case of a significant interaction, a paired-samples t-test was used for post-hoc analysis.

Step Condition	Body Weight Symmetry		Average (mm)
	Symmetrical	Asymmetrical	
One-Step (mm)	435.3(84.0)	455.0(97.7)	445.2(90.3)
Two-Steps (mm)	419.1(72.0)	395.1(69.6)	407.1(70.9)
Average (mm)	427.2(77.5)	425.1(88.9)	-----------------

Supplementary Table 6: Mean(SD) values for step width (of the first step). A 2 x 2 repeated measures ANOVA was used with stepping task (one-step vs. two-steps) and body weight symmetry (symmetrical vs. asymmetrical) as factors. In the case of a significant interaction, a paired-samples t-test was used for post-hoc analysis.

| Step Condition | Body Weight Symmetry | | Average (mm) |
	Symmetrical	Asymmetrical	
One-Step (mm)	24.4(29.8)	0.4(30.7)	12.4(32.2)
Two-Steps (mm)	29.1(30.6)	10.8(28.3)	19.9(30.5)
Average (mm)	26.7(29.8)	5.6(29.6)	-----------------

Supplementary Table 7: Mean(SD) values for the peak AP xCOM displacement after foot-contact. A 2 x 2 repeated measures ANOVA was used with step number (first vs. second) and body weight symmetry (symmetrical vs. asymmetrical) as factors. In the case of a significant interaction, a paired-samples t-test was used for post-hoc analysis.

Step Number	Body Weight Symmetry		Average (mm)
	Symmetrical	**Asymmetrical**	
First Step (mm)	430.4(40.7)	448.0(38.7)	**439.2(40.1)**
Second Step (mm)	432.3(49.2)	457.6(52.1)	**445.0(51.6)**
Average (mm)	**431.4(44.5)**	**452.8(45.5)**	-------------------

AP = anterior-posterior; xCOM = extrapolated center of mass.

Supplementary Table 8: Mean(SD) values for the peak ML xCOM displacement after foot-contact. A 2 x 2 repeated measures ANOVA was used with step number (first vs. second) and body weight symmetry (symmetrical vs. asymmetrical) as factors. In the case of a significant interaction, a paired-samples t-test was used for post-hoc analysis.

Step Number	Body Weight Symmetry		Average (mm)
	Symmetrical	**Asymmetrical**	
First Step (mm)	108.1(17.0)	104.0(19.9)	**106.1(18.4)**
Second Step (mm)	36.3(35.2)	50.6(37.9)	**43.5(36.8)**
Average (mm)	**72.2(45.5)**	**77.3(40.3)**	-------------------

ML = medio-lateral; xCOM = extrapolated center of mass.

Supplementary Table 9: Mean(SD) values for step length. A 2 x 2 repeated measures ANOVA was used with step number (first vs. second) and body weight symmetry (symmetrical vs. asymmetrical) as factors. In the case of a significant interaction, a paired-samples t-test was used for post-hoc analysis.

Step Number	Body Weight Symmetry		Average (mm)
	Symmetrical	Asymmetrical	
First Step (mm)	419.1(72.0)	395.1(69.6)	**407.1(70.8)**
Second Step (mm)	332.3(46.3)	339.1(45.4)	**335.7(45.3)**
Average (mm)	**375.7(74.1)**	**367.1(64.5)**	-------------------

Supplementary Table 10: Mean(SD) values for step width. A 2 x 2 repeated measures ANOVA was used with step number (first vs. second) and body weight symmetry (symmetrical vs. asymmetrical) as factors. In the case of a significant interaction, a paired-samples t-test was used for post-hoc analysis.

Step Number	Body Weight Symmetry		Average (mm)
	Symmetrical	**Asymmetrical**	
First Step (mm)	29.1(30.6)	10.8(28.3)	**19.9(30.5)**
Second Step (mm)	22.1(26.2)	40.0(29.7)	**31.1(29.1)**
Average (mm)	**25.6(28.3)**	**25.4(32.2)**	-------------------