EFFECT OF AGE AND FALL STATUS ON THE SCALING OF THE POSTURAL CONTROL VARIABLES TO A SCALED GAIT INITIATION TASK

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ABSTRACT

Inefficacy of the postural control system can lead to loss of equilibrium and upright stability, and ultimately a fall by older adults. Understanding the control of the center of mass (CoM) with respect to the base of support (BOS) in a progressively challenging environment that encourages individuals to scale postural control variables is of clinical importance. Subsequently, the long-term goal of this work is to optimize evidence-based balance rehabilitation programs for older adults with balance impairments. However, it is currently unclear how older adults perform scaled tasks during gait initiation (GI). It is possible that older adults with a fall history would have problems scaling postural control variables in response to scaled postural challenges. The inability to scale postural control variables could further increase their chance of becoming unstable. Therefore, the purpose of this dissertation was to understand how older adults control balance in postural challenging GI tasks such as scaled step distance and step width tasks. The scaled step distance task involved stepping over an obstacle placed at progressive distances from an individual's body. The scaled step width task involved stepping over an obstacle and stepping on medial or lateral targets placed on the floor.

In the first study, we observed healthy adults (n = 17 with a mean age of 27.8 ± 6.3 years) performing scaled GI tasks in the presence of an obstacle in the sagittal plane. Overall, healthy adults increased anterior-posterior (AP) center of mass-center of pressure (CoM-CoP) displacement and decreased trunk forward lean at maximum CoM-CoP displacement during swing in response to scaled GI tasks. As a part of an exploratory

analysis, we also found that the variability of AP CoM-CoP displacement at maximum CoM-CoP displacement during swing did not change, but the variability of trunk lean at maximum CoM-CoP displacement during swing increased. The findings suggest that healthy adults scale the AP CoM-CoP displacement and the AP trunk lean per the task demand.

For our second study, we compared the scaling behavior of postural control variables in the sagittal plane between older non-fallers (n = 12 with a mean age of 73.41 ± 3.91 years) and fallers (n = 12 with a mean age of 71.66 ± 5.53 years) while performing scaled GI tasks in the presence of an obstacle in the sagittal plane. In this study, we found that older fallers reduced AP CoM-CoP displacement prior to the toe-off compared to non-fallers. However, at maximum CoM-CoP displacement during swing, AP CoM-CoP displacement was not different between the groups. We also found that older fallers increased peak trunk forward lean during swing compared to non-fallers. These findings suggest that older fallers adopt a greater conservative approach with AP CoM-CoP displacement than non-fallers prior to the toe-off of the scaled GI task to compensate for instabilities that may arise due to the peak AP trunk lean in the swing phase.

In our third study, we compared scaling differences between older non-fallers (n = 12 with a mean age of 73.41 ± 3.91 years) and fallers (n = 8 with a mean age of 70.62 ± 5.26 years) during scaled step width GI tasks in the frontal plane. We found that older fallers reduced medial-lateral (ML) CoM-CoP displacement at maximum CoM-CoP displacement during swing compared to non-fallers. However, the groups were not different prior to the toe-off of the GI tasks. We also noted that older non-fallers increased trunk tilt toward the stance limb more in the obstacle than the no obstacle condition irrespective of step width conditions, whereas fallers had increased trunk tilt toward the stance limb irrespective of the obstacle presence or step width conditions. These findings suggest that older fallers adopt a greater conservative approach for the ML CoM-CoP displacement and ML peak trunk tilt than non-fallers during the swing phase of scaled GI tasks for scaled step width conditions.

This study provides a framework for investigating postural stability in older adults with clinical problems in a laboratory setting. Looking at a broader comparison between healthy adults (less than 40 years of age) and older adults (above 65 years of age) with the scaled step distance tasks during GI, it seems that older adults prefer a cautious strategy based on CoM-CoP displacement at maximum CoM-CoP displacement during swing than healthy adults. Furthermore, young adults and older adults, whether non-fallers or fallers, approach in a way that they can complete the task successfully while ensuring safety. The results of these investigations show that changes in the CoM-CoP displacement are dependent on the phase of the GI tasks and plane in which the variable was examined. This study's findings may be used to identify postural instability and design targeted balance rehabilitation protocol in balance-impaired populations, such as older adults, children with cerebral palsy, and other neurological populations.

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CHAPTER 1

LITERATURE REVIEW

Importance of postural stability

Postural stability is important for movement control of everyday functional activities. Postural instability results from improper center of mass (CoM) control with respect to the base of support (BOS), which can lead to loss of balance or falls [1]. One in every four older adults aged $\geq$ 65 years fall each year in the US, and, with increasing age, fall incidences in old adults increase [2]. In 2015, the total cost of non-fatal injuries caused by fall was estimated to be $31.3 billion, adjusted for inflation [3]. The total direct and indirect medical cost of fall-related injuries such as a head injury or a hip fracture is estimated at $54.9 billion by 2020 [4], a significant burden to the health care system.

Previous studies have suggested that static and dynamic balance control is reduced in older fallers compared to older non-fallers [5–7]. Two common causes of falls by older adults are related to transfers and trips [8,9]. Insufficient transfer of body weight when initiating gait can result from a loss of control of the CoM with respect to the BOS while trips result from improper control of the swing limb while advancing the BOS [8]. Such inappropriate weight transfer in the frontal plane and tripping might be associated with poor postural control in older adults. Postural instability has been found to increase with increasing difficulty of the task and with increasing age [10], which suggests that in older adults, postural stability can be challenged with increasing task difficulties to investigate postural control. Older adults may actually increase their dynamic stability compared to young adults when challenged with tasks of increasing difficulty as a strategy to increase control on CoM kinematics [11].

Gait initiation (GI) and postural stability

GI is a common functional task that requires one to transition from a stable BOS (i.e., standing), to a continually changing BOS, (i.e., gait) by shifting weight towards the swing limb and then back to what will be the stance leg during GI. The transfer of weight from the upcoming swing limb to the stance limb in the anticipatory postural adjustment (APA) phase, along with a backward shift of the center of pressure (CoP), helps generate forward momentum to initiate the step in the swing phase during GI [12]. Transfers rely on sufficient control of the CoM and its regulation with the CoP. Moreover, the separation between CoM and CoP is critical to initiate gait [13], and this separation is termed CoM-CoP displacement. GI can be challenging if the CoP shift is not sufficient to increase the acceleration of the CoM in the direction of movement. It has been shown in a previous study that during GI, older adults reduce CoP posterior shift in the APA phase compared to young adults, which reduces forward momentum [12]. Similarly, during GI, older fallers reduce the backward CoP shift in the APA phase compared to older non-fallers that limits forward movement of the CoM and increases stability at the toe-off [14]. Given this, understanding postural stability in older fallers during GI becomes important.

In older fallers, it has been found that during GI, the first step length was reduced compared to non-fallers, and step length variability was higher [15]. It is also reported that step length and step width variability during GI is greater in older adults than in young adults [16]. Moreover, older adults have been shown to reduce forward-directed CoM velocity when performing a wider step width task compared to young adults [11]. Based on this, changes in step length and step width could indicate changes in dynamic stability with aging and fall status.

Scaling and postural stability

In the transition from GI to steady-state walking, scaling postural control variables to the task constraints is critical to maintaining stability. Scaling of task performance has not been directly defined by previous studies; however, researchers have used the concept of scaling task performance based on scaled tasks. For example, Park et al. (2004) inferred scaling as a function of increased postural perturbation (by backward translation of the increasing magnitude of the support surface) and observed the change in postural responses (hip vs. ankle joint trajectories, angular accelerations, and torques) while recovering initial upright posture [17]. Another study by Adkin et al. (2000) applied the concept of scaling to an increased level of postural threat (increased surface heights) while standing quietly and observed the change in postural responses (a linear decrease in CoP displacement amplitude) [18]. Hurt et al. (2015) reported scaled task difficulty led responses such as manipulation of hip abduction moment, the velocity of CoM, lateral CoM displacement, and step time while performing lateral stepping task compared to normal walking. Additionally, increasing task difficulty (from short to medium to long stepping targets) compared to normal walking also led to larger changes in peak CoM velocity, suggesting scaling of the task performance with the size of the step [11]. Another study by Yiou et al. (2016) applied scaling term to task of increasing obstacle height during GI in young adults and observed significant changes in medial-lateral (ML) peak CoP shift and ML APA duration[19]. Thus, a scaled task-induced scaled task performance could be useful in understanding or uncovering any underlying reduction in CoM control that could better identify issues of postural stability than just observing individuals under more simplified conditions. Furthermore, it can be inferred that the control of CoM in response to scaled tasks can result in a more conservative approach that may require greater imposed control over the postural control variables. For this dissertation, scaling or scalability is defined as changes in task difficulty that lead to significant differences in task performance from one trial to another. Importantly, scaling of task performance is one way to understand how the nervous system prioritizes certain variables to ensure stability in task performance.

Postural stability is defined as the maintenance of control of the CoM within the base of support. Strategies adapted to control CoM and increase stability involve modifying stepping by changing step length or step width and altering trunk motion [20]. Scaled step distance and step width tasks are measures that can be used to challenge postural stability [11,21]. Also, stepping over obstacles while performing such tasks can further increase challenges to maintain upright balance. A previous study has illustrated that during GI and stepping over an obstacle of increasing height, healthy young adults scale by reducing the peak AP CoP displacement at the toe-off, which is a well-established marker of postural stability, to increasing postural challenges imposed by obstacle heights [19]. Thus, during a performance of increasingly challenging tasks, this reflects a tighter control by the central nervous system (CNS) to regulate the CoM movement within the base of support, as the difference between CoP and CoM is directly proportional to CoM acceleration.

CoM-CoP scaling and postural stability

Previous research has suggested that in response to increasing obstacle height, healthy young and older adults and individuals with Parkinson's disease scale anterior-posterior (AP) CoM-CoP displacement to maintain postural stability [19,22,23]. Scaling the step width from preferred to narrow has been shown to reduce CoM displacement and margin of stability in both young and older adults [24]. Scaling of dynamic stability with scaled step length from shorter to longer steps has also been shown during walking in young adults [21]. Given this, it indicates that inability to scale CoM-CoP displacement (one of the postural control variables) to the challenging tasks may lead to postural imbalance. Furthermore, to the best of my knowledge, only one study looked at stability during GI when performing a scaled step distance task [19]. In this paper, the authors looked at some postural control variables in young adults while they stepped over an obstacle placed at different distances during APA, foot-off, and heel-strike. Among the results reported in that study, there was a lack of details on postural control variables such as CoM-CoP displacement, trunk angle, CoM velocity during GI while performing scaled step distance or step width tasks.

It is suggested that the larger the CoM-CoP displacement individuals can keep in controlled situations, the greater the dynamic balance control ability shown by them [25]. This suggests that individuals who maintain greater CoM-CoP displacement while performing GI have better balance control than those with a decreased CoM-CoP displacement. During GI, it is also evident that CoM-CoP displacement reaches its maximum just prior to leading limb foot strike [25]; it makes this time point essential to study to understand postural stability behavior. Moreover, the CoM-CoP displacement during GI is also considered a sensitive marker of dynamic stability in balance impaired individuals [26], which suggests that CoM-CoP displacement should be further investigated using different postural stability challenging tasks to investigate balance control.

While normal GI requires separation of the CoM and the CoP as shown in Figure 1, it can be postulated that stepping over obstacles at progressively longer distances or varying step widths during GI will challenge CoM control and hence require a certain CoM-CoP displacement to avoid becoming unstable. There is evidence that the CoM-CoP displacement at various events during GI is larger in young adults, implying better postural stability in young compared to older adults and individuals with Parkinson's disease, as illustrated in Figure 2 [27]. Peak CoM-CoP in APA, CoM-CoP at toe-off, and CoM-CoP in the swing phase while performing different challenging tasks such as stepping over an obstacle at increasing distances during GI can give us insight into how the CoM and CoP are regulated to keep the body stable during perturbation. Importantly, this has not been reported in detail in young adults, older non-fallers, and fallers. In GI, peak CoM-CoP displacement during the end of the swing limb heel strike was smaller in more versus less balance impaired individuals [25].

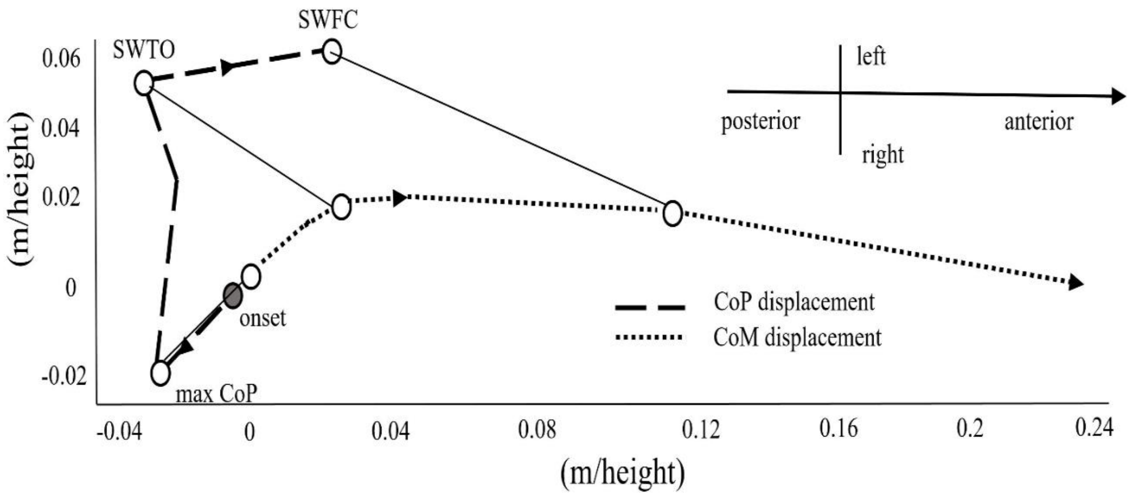

Figure 1. The separation between CoM and CoP (indicated by the solid lines) at different events of GI. maxCOP, the maximum shift of the center of pressure; SWTO, swing limb toe-off; SWFC, swing limb foot contact. Adapted from Mickelborough et al. 2004 [28]

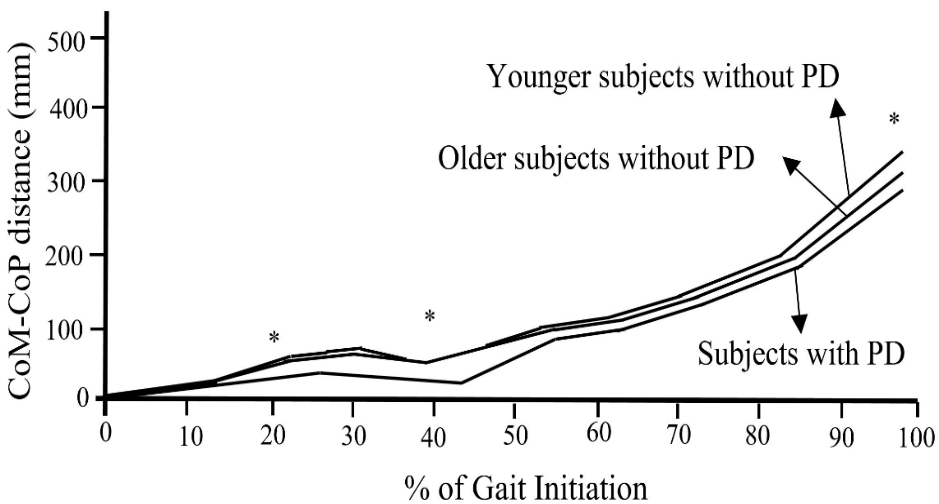

Figure 2. CoM-CoP distance for various population. * represents a significant difference between the population at different phases of GI. Adapted from Martin et al. 2002 [27]

Trunk control

Trunk position plays an important role in CoM positioning relative to the BOS as most of the total body weight is located in the trunk segment. Trunk angular displacement has also been shown to be a valid measure to assess balance in older adults [29,30]. During GI, it has been shown that in the sagittal plane, the trunk tilts forward in young adults as displayed in Figure 3 [31–34]. In the frontal plane, the trunk leans toward the stance limb in the APA phase and then leans toward the swing limb in the swing phase [34]. Imbalance in AP direction or ML direction caused by tripping over an obstacle or stepping onto certain step widths involves trunk motion anteriorly/laterally, and hence proper trunk control to keep CoM over the BOS is required for stability.

Trunk control has been suggested to be important to walking [35] and recovering from postural disturbances [36]. Despite this, few studies have investigated trunk control during GI tasks. A previous study has also reported that trunk kinematics are related to step width in both young and older adults while walking [35]. Trunk kinematics have also been studied during different walking speeds in older adults, where it was found that older women had greater sway of the trunk than older men [37]. Moreover, the trunk motion is not changed significantly in older fallers compared to non-fallers when walking speed is changed [38]. However, in another study, average trunk flexion during walking between older fallers and non-fallers was similar in these groups [39]. Older adults have also shown an increased trunk range of motions in the frontal, sagittal, and transverse planes to navigate an obstacle course during walking to control CoM, which might affect their dynamic stability and make them more unstable as presented in Figure 4 [40]. Although it is known that trunk motions are affected when stepping over obstacles in older adults [40,41], a significant gap exists regarding knowledge on trunk kinematics while performing scaled step tasks using obstacles as spatio-temporal constraint during GI.

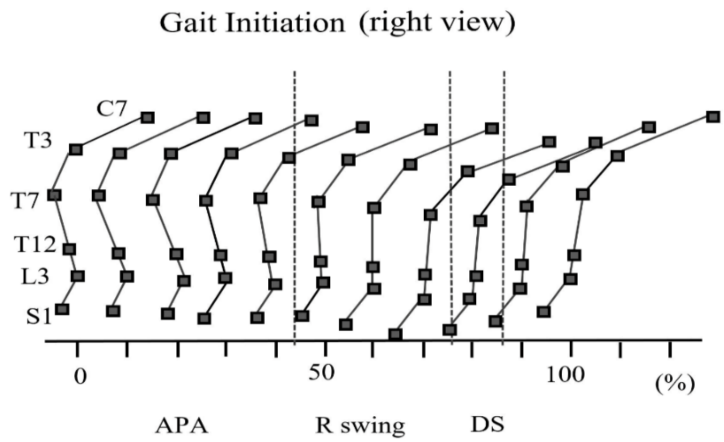

Figure 3. Trunk lean in the sagittal view during GI. Adapted from Ceccato et al. 2009 [34]

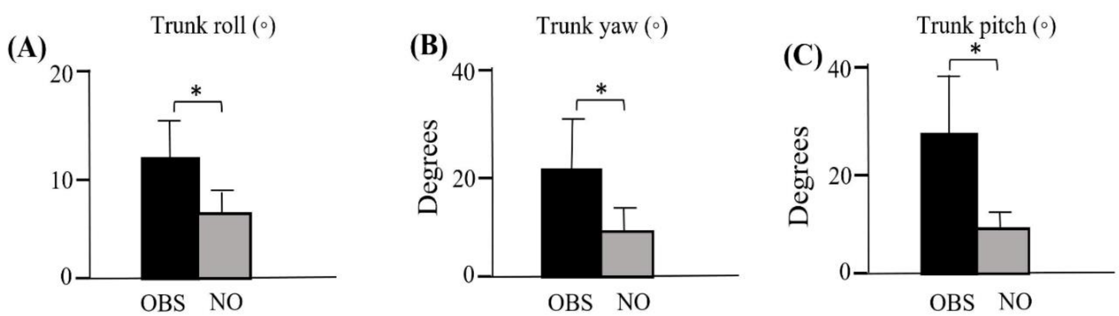

Figure 4. Trunk range of motions in the frontal (A), transverse (B), and sagittal plane (C). White and black bars represent two obstacle conditions, and grey bars represent no obstacle condition. * show a significant difference between obstacle and no obstacle conditions. Adapted from Lowrey et al. 2007 [40]

CoM velocity control

In addition to angular or positional data, velocity data also plays an important role in stability. There is not much known on CoM velocity during scaled step tasks during GI, however, significant literature exists on CoM velocity during walking. CoM velocity is used to derive the margin of stability [42], which is another marker commonly used to investigate the stability of individuals. Moreover, CoM velocity is also suggested to be predictive of foot placement location during walking [43], which is critical to regaining stability after perturbation. During walking, AP CoM velocity at toe-off is decreased in older fallers compared to older non-fallers and young adults [42].

Interestingly with limited knowledge on CoM velocity during GI, it has been reported that in healthy young adults, the AP velocity of the CoM at foot off is increased for stepping over obstacles at the longer distance during GI [19]. The velocity of the CoM at the heel strike during GI also needs to be controlled to avoid instability. Notably, the AP velocity of the CoM increases with the progression of GI and remains elevated at heel strike in young adults [13,44–46].

ML CoM velocity at the heel strike in GI is also high in young adults compared to older adults [46]. However, the ML velocity of CoM at foot contact was not different in young adults while stepping over obstacles at a longer distance during GI [19]. ML CoM velocity, at foot contact, if increased, requires either an increased hip abductor moment [11] or a compensatory step to avoid becoming unstable. During GI, the AP and ML velocity of the CoM at heel strike [47] and the velocity of the CoM at toe-off in older adults are lower than in young adults [48]. Interestingly, peak ML velocity of the CoM while stepping over obstacles at different heights during walking was reported to be greater in older fallers compared to non-fallers [22]. However, there is a lack of information on CoM velocity in older adults at greater risk of falling during GI on scaled stepping tasks.

Stepping over obstacles and postural stability

Biomechanical analysis of stepping over obstacles, another common daily task, has been used to assess balance and gait function in older adults [49,50]. This task challenges postural stability and helps investigate deficits in the postural control and gait systems. Compared to stepping without an obstacle, stepping over obstacles require that individuals ensure proper clearance of the swing limb over the obstacle to avoid a trip while also increasing the time spent on the stance limb, which provides a challenge to balance. In older adults, stepping over obstacles during GI results in the increased step time compared to typical GI. This finding suggests that older adults use a more cautious behavioral strategy to ensure proper clearance of the obstacle while maintaining balance in the execution of the task [51]. Given the above findings, the combination of GI and stepping over obstacles could impose postural control challenges in older adults with greater fall risk and hence can be used as a task to investigate balance and gait functions.

AP and ML postural stability

For individuals identified as fallers, AP and ML stability are affected due to poor postural control while performing various tasks. In GI, as AP and ML stability are required to progress forward and shift weight from one limb to the other, older adults in general and fallers specifically demonstrate decreased hip abductor strength, which could affect task performance and how they maintain AP and ML stability when they perform AP scaled and ML scaled tasks during GI. The CoM-CoP displacement at the heel strike in the sagittal plane is greater in non-fallers versus fallers during level walking but not different in ML direction [52]. Moreover, while stepping over obstacles of different heights during level walking, maximum CoM-CoP displacement in AP direction decreased in older adults as opposed to young adults, but there was no difference in maximum CoM-CoP displacement in ML direction [41] signifying there are differences in postural stability control in different anatomical planes.

Frontal plane stability during gait is critical in older adults as instabilities in ML direction can lead to a lateral fall, which increases the risk of a hip fracture six times over falls forward or backward [53]. However, in a study, older adults' prioritization of sagittal plane stability over the frontal plane stability when performing voluntary lateral step may be to regain balance [54]. During a voluntary lateral step initiation, prior to toe-off, older adults show two postural adjustments (loading and unloading) in the swing limb compared to young adults who show a single postural adjustment, i.e., unloading [55]. These adjustments are based on vertical ground reaction forces. The swing leg is initially loaded (increasing vertical ground reaction forces), to progress the CoM toward the stance leg. It is followed by decreasing vertical ground reaction forces so that the swing leg is unloaded for the next step. The above-mentioned increase in postural adjustments is reflective of proper CoM control requirements in older adults. Especially in older adults with reduced postural stability, proper CoM control becomes critical when changing step width during GI.

ML foot placement has been considered as one of the mechanisms for ML stability control [56] and is reduced with age [57–59]. There is evidence that during voluntary step initiation, older adults compared to young adults showed reduced peak lateral force under the stance limb when perturbed with an unexpected stepping target sideways to maintain upright stability [60]. Older adults, stepping to laterally directed targets during walking, demonstrated a decreased hip abductor moment compared to young adults [11]. In young adults, increasing gait speed increased lateral CoM displacement at heel contact during GI, which was associated with an increased step width to presumably maintain ML dynamic stability [61]. Modifying AP and ML foot placement will stress the sagittal and frontal stability control in older adults and, in theory, require modification to their APA and swing phase. Thus, scaling a GI task can help discriminate postural control variables or balance control variables such as CoM-CoP displacement, CoM velocity, and trunk angle, which play a critical role in CoM control between older adults with greater fall risk and healthy older adults.

Summary of Gaps in the Literature

Based on the above-reviewed knowledge, evidence supports that postural control variables such as CoM-CoP displacement, trunk motion, and CoM velocity during scaled GI tasks. However, there is a lack of information on the scaling of CoM-CoP displacement, trunk kinematics, and CoM velocity during scaled GI tasks, especially in older non-fallers and fallers. Thus, it is important to investigate behaviors used to maintain postural stability in older non-fallers and fallers during scaled step distance and step width GI tasks to better understand the underlying differences between them in terms of postural control ability.

Notably, understanding of scaling of CoM-CoP displacement, trunk kinematics, and CoM velocity during scaled GI tasks will provide us an insight into the control of CoM kinematics while performing tasks of increasing difficulty.

Specific Aims

The long-term goal of this study is to optimize the evidence-based balance rehabilitation program for older adults. The overall objective of our study is to understand how older adults control balance in postural challenging GI tasks. Our central hypothesis is that when challenged with scaled obstacle distances and scaled step widths during GI, older adults with balance impairment will not be able to scale postural control variables properly compared to healthy older adults. The rationale for this study is that it will provide a better understanding of postural stability in healthy young adults, healthy older adults, and older adults with balance impairment. To test the central hypothesis, the dissertation will utilize two experiments: 1) challenge balance using scaled obstacle distances stepping task during GI 2) challenge balance in stepping task during GI by modifying step widths. Using these experiments, we propose the following three Specific Aims:

Aim 1) To characterize postural control in a scaled GI task in the presence of spatio-temporal constraints imposed by an obstacle in healthy adults.

The study will use increasing obstacle distances to step over during GI in healthy adults. The study will help determine how healthy adults scale balance control variables to increasing obstacle distances during GI.

Hypothesis: In response to increased step distance when crossing obstacles, healthy adults will scale the CoM-CoP displacement with increasing postural disturbance during GI.

Rationale: Scaling of the postural control variables in healthy young adults will allow us to explore the characteristics of a stable postural behavior. The data obtained from Aim 1 will provide us the framework to study postural stability in older fallers and feasibility to conduct in a laboratory setting with participants.

Aim 2) To compare the scaling of the postural control variables between older non-fallers and older fallers to a scaled step distance task during GI in the presence of spatio-temporal constraints imposed by an obstacle.

The study will use increasing obstacle distances to step over during GI to determine how older fallers scale balance control variables to increasing obstacle distances during GI compared to older non-fallers.

Hypothesis: During GI, compared to non-fallers, older fallers will not scale their posture control, and that the fallers will not show scaling for the increased distances while performing progressive stepping distances resulting in group x distance interaction of 1) peak AP CoM-CoP displacement during APA and 2) AP CoM-CoP displacement at the toe-off and, 3) AP CoM-CoP displacement at maximum CoM-CoP displacement during swing.

Rationale**:** Previous research provides evidence that peak CoM-CoP displacement at the heel strike can discriminate between older fallers and non-fallers during GI [25]. It is reduced in older adults with balance impairment versus healthy older adults [24,26,60].

However, how exactly postural control variables differ in older fallers versus older non-fallers with increased postural challenges such as in a scaled step distance task during GI is not known.

Aim 3) To determine the extent to which older fallers and non-fallers scale the postural control variables when stepping over an obstacle while increasing or decreasing step width to challenge ML stability**.**

The study will use the task of stepping over an obstacle during GI along with increased or decreased step width to determine how older fallers scale postural control variables to scaled step widths during GI compared to older non-fallers.

Hypothesis: Older fallers performing a GI task with targeted initial step widths would result in no scaling of 1) peak ML CoM-CoP displacement during APA and 2) ML CoM-CoP displacement at the toe-off and, 3) ML CoM-CoP displacement at maximum CoM-CoP displacement during swing compared to non-fallers resulting in group x step distance x step width interaction.

Rationale: Balance control measures primarily related to ML postural stability is increased more in older fallers than non-fallers [62]. Asking older fallers to step over obstacles while modifying step width will challenge their postural stability system, particularly in the frontal plane, and a reduced capacity to control CoM may result in a loss of balance while performing the task. Variation in step width has also been illustrated to be associated with variation in trunk motions in healthy older adults during walking [35]. Older adults, while stepping over an obstacle during walking, have been shown to reduce step width [39]. Notably, extreme step width variability during walking at normal speed has been reported in older adults with a history of falls [21]. Based on the above details, changing the step width in this aim can unfold the nature of ML stability in more or less stable older adults.

Summary

Completion of this project will help delineate strategies used to maintain postural stability by older fallers and non-fallers when performing scaled step tasks during GI. It will further aid in the investigation of postural instability in older fallers with difficult task performances. Data in the AP and ML direction with the AP and ML tasks respectively will help understand the importance of postural stability in those specific directions for older adults, especially fallers. This project design can be used as a framework to investigate postural imbalance in other clinical populations with gait issues. Moreover, the results of this project can lead to the development of a novel rehabilitation-based targeted intervention for balance-impaired individuals in the future. Overall, this project can inform improvement in mobility and quality of life in older individuals.

CHAPTER 2

CHARACTERIZATION OF BALANCE CONTROL IN A SCALED STEP DISTANCE TASK DURING GAIT INITIATION IN HEALTHY ADULTS

Introduction

Safe mobility in a complex environment requires that the nervous system flexibly modify balance parameters to ensure the maintenance of upright stability. Upright stability can be achieved via the scaling of postural control variables. Scalable locomotor task performance may be required to step over undesirable step locations (e.g., water puddles), or avoid obstacles (e.g., door thresholds). When these tasks are initiated from a quiet stance, the typical pattern of gait initiation (GI) results in a transfer of weight toward the swing/leading limb and back to the stance limb to prepare for the step, which can be affected by the size of the initial step to be taken [63]. Such transfers during GI are important to help generate forward momentum and ensure a stable transition from quiet standing to walking [12].

In general, GI can be divided into the anticipatory postural adjustment (APA) phase and the swing phase. The interaction between the center of pressure (CoP) and the center of mass (CoM) during GI regulates the CoM trajectory [13] (Figure 1. from the previous chapter). During GI, the CoM moves toward the stance limb and forward, and the CoP shifts toward and backward under the leading limb in the APA phase. The CoM then shifts towards the leading limb and forward while the CoP shifts to the trailing limb and forward during the swing phase of GI. All of these movements [31,64–66] create separation between CoM and CoP. This separation between CoM and CoP is termed as CoM-CoP displacement. CoM-CoP displacement reaches its maximum immediately prior to the leading limb foot strike during the swing phase [22], making it an interesting time point to study postural stability during scaled step distance tasks.

Peak CoM-CoP displacement is shown to be a sensitive marker of stability in older adults as the peak CoM-CoP displacement was significantly reduced in balance impaired older individuals compared to healthy older adults [67]. Individuals with balance impairment minimize CoM-CoP displacement compared to individuals with no balance impairment, suggesting that greater CoM-CoP displacement may relate to better dynamic balance control ability [25]. Moreover, individuals with balance impairment may exhibit a more cautious strategy due to fear of falling or perceptual deficits that results in a lower CoM-CoP displacement relative to individuals without a balance impairment. Further, CoM-CoP displacement may be the critical measure regulated by the nervous system when performing scaled step distance tasks, even in young adults.

Postural instabilities can arise by insufficient anticipatory postural adjustments and inappropriate stepping when initiating gait. Modulations in the APA phase of GI help gain postural stability during the swing phase when performing tasks in constrained environments [68]. Stepping over an obstacle requires body weight to be supported on the stance limb for a prolonged duration, requiring the nervous system to exert greater postural control to maintain balance throughout the task. To ensure the initiation and maintenance of safe mobility, healthy adults can scale postural responses relative to the postural threat [18,69,70] to regulate the CoM trajectory with respect to the base of support. Although

CoM-CoP displacement has been reported while stepping over obstacles of different heights and distances [19,22], no investigations have quantified how healthy adults scale CoM-CoP displacement in response to a scaled step distance task in the presence of obstacles during the swing phase of GI. Studying the scaling of CoM-CoP displacement to scaled step distance tasks during GI in healthy adults could help to provide a reference to compare and understand the scaling in the clinical population.

While stepping, controlling the trunk is crucial due to its large segmental mass and vertical height off the ground. Two-thirds of a person's body weight is located in the head arms and trunk, and this segment's CoM is located approximately two-thirds of a person's height off the ground. During GI, the trunk reaches maximum forward lean at the leading limb foot strike in healthy adults [34] to maintain forward momentum needed to move forward. Healthy adults demonstrate variability in anterior-posterior (AP) trunk control (i.e., forward lean versus backward lean) during GI, and hence utilizing different strategies [71], which could result in maintenance of postural stability. Previous research has shown that variations in trunk kinematics are related to variations in step kinematics to ensure gait stability during walking [35]. These findings suggest trunk motion could play a crucial role in postural stability in a scaled step distance task during GI as well. Nevertheless, minimal information exists on trunk lean during GI in scalable task constrained by obstacles in healthy adults.

Scaling of CoM-CoP displacement and trunk angle in the swing phase in response to a scaled step distance task in the presence of obstacles during GI can provide insight into how the healthy nervous system ensures postural stability while performing these tasks.

Scaling or scalability in this study is defined as increases of task difficulty such as increases of the obstacle distance that lead to significant differences in task performance such as amplitude of CoM-CoP displacements from one trial to another. Variability in task performance during GI, as characterized by such measures, suggest alterations in the neural control of GI. The impact of the scaled step distance and obstacle task on the amount of variability of CoM-CoP displacement and trunk stability is also unknown. Changes in variability could provide us insight into neural control of GI. Therefore, gathering information about the variability of CoM-CoP displacement and trunk motion during a scalable task will enhance the understanding of neural control during GI in healthy adults. Thus, the purpose of this research was to characterize postural control in a scaled GI task in the presence of spatio-temporal constraints imposed by an obstacle in healthy adults. We hypothesized that in response to increased step distance when crossing obstacles, healthy adults will scale the CoM-CoP displacement with increasing postural disturbance during GI.

Materials and methods

Participants

Nineteen healthy college students consented to participate in this study. Each participant signed the University's Institutional Review Board (IRB) approved consent form (IRB## 22499). All participants reported no history of neurological conditions and were independent walkers. We were missing the pelvis markers location in one subject during data collection, and the other subject was making a toe strike instead of heel strike during the trials for stepping over obstacles condition, which made identification of the first step incalculable. Therefore, we were able to collect full sets of data on 17 participants (10 females, 7 males with a mean age of 27.8 (±6.3) years, mean height of 1.7 (±0.09) m, and mean weight of 72.56 (±13.2) kg).

Experimental setup

In this cross-sectional study, we asked participants to stand on two separate force plates that were embedded flush with the floor, with feet placed at a comfortable width. The initial foot position was marked on the force plates to standardize foot position for all trials. Participants wore their shoes during the data collection. We instructed them "when you hear 'GO' step forward with your right leg first, over the obstacle if there is one, and walk forward 3-4 steps and then return to your starting position". Perform all the trials at your comfortable speed, and you can look straight ahead or, if needed, look down to perform the trials". The scaled stepping task included three conditions of difficulty: no obstacle (NO), near obstacle distance (NEAR), and far obstacle distance (FAR). NEAR and FAR conditions were defined as an obstacle placed at 25% and 65% of the individual's leg length, respectively. These distances were based on our pilot sample, where we found that, on average, ~80% of the individual's leg length was the maximum distance that individuals were able to cross over the obstacle successfully. We used a 17 cm, similar to a curb height, high wooden bar as the obstacle, shown in Figure 5. We marked NEAR and FAR on the floor from an individual's toe starting position. For NEAR and FAR conditions, we placed the obstacle at one of the marked distances before the beginning of the trials. Before the start of the trial, participants were able to see the walkway and the obstacle if placed in front of them. There were 50 trials in each condition. A total of 150 trials were performed in a randomized order. In cases where a participant hit or missed the obstacle, a repeat trial was done. Participants were given a mandatory rest of 2-5 minutes after each block of 50 trials. A spotter was present for all trials for subject safety.

Figure 5. Participant performing stepping over the 17 cm high obstacle for the experiment

We used a Plug-in-gait marker set to capture kinematics using an 11-camera motion analysis system (Qualysis, Inc) sampling at 120 Hz and tracked offline using commercial software (Qualisys track manager system). Anthropometric measures were collected using caliper of ankle, knee, wrist, hand, and shoulder widths for input into the kinematic model. We captured kinetic data using 2 embedded AMTI force plates (Advanced Mechanical

Technology Inc., Newton, MA) sampling at 200 Hz. The synchronized kinematic and force plate data were exported for analysis.

Data analysis

For this investigation, the AP CoM-CoP displacement and trunk lean were used as the primary measures of postural control. We estimated the whole body CoM position from reflective markers based on a 12-segment rigid body model. To calculate whole body CoP, we combined CoP data from two force plates. The AP CoM-CoP displacement was calculated as the displacement between CoM and CoP position in the sagittal plane. Using the C7 marker position and mid-PSIS location, the absolute trunk angle in the sagittal plane was calculated. The custom code in MATLAB 2018a software was used for these calculations. To filter kinematic and kinetic data, we used a 4^{th} order low-pass Butterworth filter with a cutoff frequency of 10 Hz and 20 Hz, respectively. For each trial, we extracted data of the 1^{st} step up to the right foot heel strike. Within that 1^{st} step, toe-off of the right foot was noted as the time the vertical ground reaction force under the right foot dropped below a threshold of 60N, and right heel-strike as the local minima in the vertical plane of the right heel marker. To mark the heel strike data, we used minimum right heel velocity in the y-direction. To get data for the time at maximum CoM-CoP displacement during swing, we extracted data from a time point where CoM-CoP displacement reaches maximum. For each participant, the mean and standard deviation of the data across trials were analyzed in each condition.

Statistical analysis

For our analyses, we used a one-way repeated measures ANOVA with Bonferroni correction for post-hoc to compare means between different conditions for both AP CoM-CoP displacement (m) and trunk angle (degrees) at maximum CoM-CoP displacement during swing, step time (s), and step length (m). A one-way repeated measures ANOVA compared variability between different conditions for both AP CoM-CoP displacement and trunk angle at maximum CoM-CoP displacement during swing. Standard deviation (SD) quantified variability. Further, as a part of exploratory analysis, Pearson product-moment correlation coefficients (*r*) examined the relationship between (1) AP CoM-CoP displacement and trunk angle; and (2) SD of AP CoM-CoP displacement and SD of AP trunk angle at maximum CoM-CoP displacement during swing in each condition. Correlation coefficients analyzed the relationship between these two variables: (1) across subjects and condition; and (2) within condition but across subjects. Alpha was set at 0.05 for all the analyses using SPSS software.

Results

Participant characteristics

In this study, we had to remove data of 2 participants from the analysis, as mentioned in the methods section. We ran the analysis on 17/19 participants. The characteristics of the participants are presented in Table 1.

Table 1. Participant characteristics

N	Gender (M/F)		Age (y)	Mass (kg)	Height (m)	NEAR Obstacle Condition (m)	FAR Obstacle Condition (m)
17	7/10	Mean	27.8	72.56	1.7	0.21	0.55
		SD	6.3	13.2	0.09	0.01	0.02

Effect of task constraints on step length and step time

Both step length and step time increased with the scaling task (Table 2, $p<0.001$ for both). In the post-hoc analysis, step length was significantly greater in 1) FAR versus NEAR and FAR versus NO ($p < 0.001$) and 2) NEAR versus NO ($p < 0.001$) showing increases in step length with increasing obstacle distances. We also observed that step time was greater in 1) FAR than NEAR ($p = 0.007$) 2) NEAR than NO ($p < 0.001$), and 3) FAR than NO ($p < 0.001$) showing that the time required to take a step over the obstacle increases with the obstacle distance.

Table 2. Step length and step time for all the conditions (mean ± SD)

Obstacle Condition	Step length (m)	Step time (s)
NO	$0.64 \pm 0.05^{*,\#}$	$0.47 \pm 0.06^{*,\#}$
NEAR	$0.73 \pm 0.05^{**}$	$0.75 \pm 0.11^{**}$
FAR	0.91 ± 0.05	0.80 ± 0.12

*denotes significance between NO-NEAR, #denotes significance between NO-FAR, and **denotes significance between NEAR-FAR

Scaling of CoM-CoP displacement and trunk angle at maximum CoM-CoP displacement during swing

The CoM-CoP displacement increased and the trunk forward lean angle decreased with the scaling task (Figure 6). Specifically, the CoM-CoP displacement was greater in FAR than NEAR than NO ($p < 0.001$) while the trunk angle forward lean was lower in 1) FAR than NEAR ($p < 0.001$) 2) FAR than NO ($p < 0.001$), and 3) NEAR than NO ($p = 0.002$). For the AP CoM-CoP displacement during swing, an example of individual data points is shown in Figure 7. For AP trunk angle during swing, an example of individual data points is shown in Figure 8.

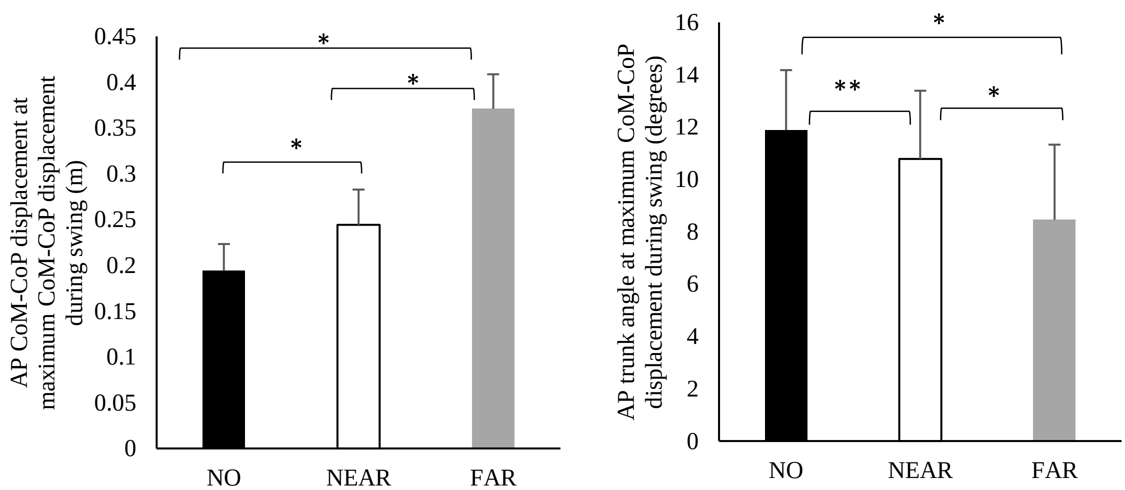

Figure 6. AP CoM-CoP displacement and trunk angle at maximum CoM-CoP displacement during swing within the conditions (mean ± SD). Positive values in trunk angle graph shows forward trunk lean. * significant difference ($p < 0.001$) . ** significant difference ($p = 0.002$)

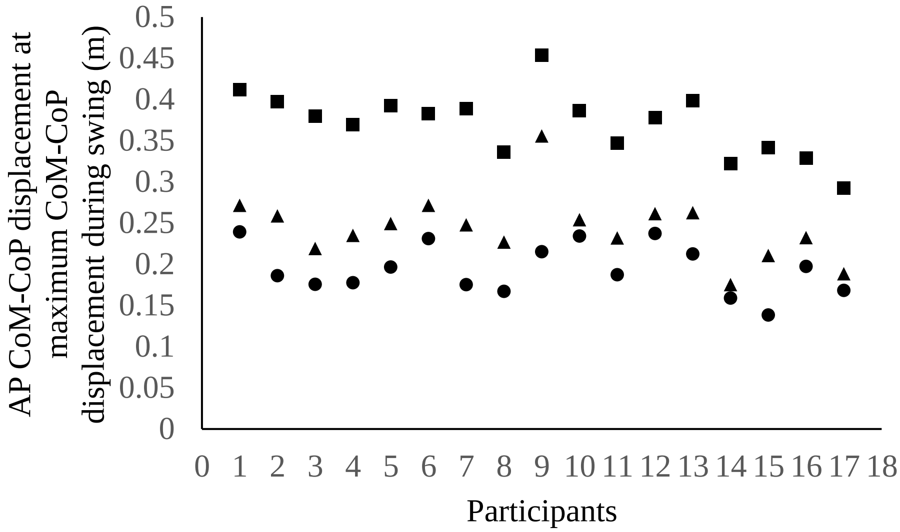

Figure 7. Individual data points for AP CoM-CoP displacement during swing for all the 17 participants. Circles represent NO condition, triangles represent NEAR condition, and squares represent FAR condition.

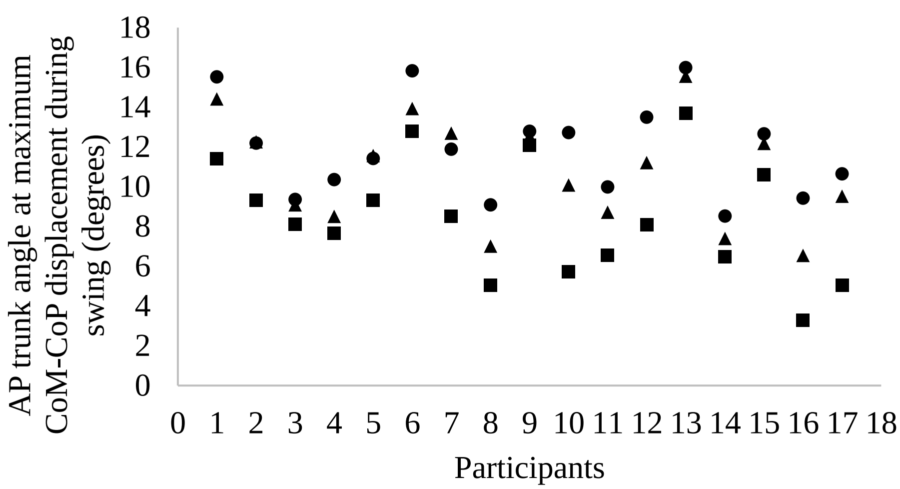

Figure 8. Individual data points for AP trunk angle during swing for all the 17 participants. Circles represent NO condition, triangles represent NEAR condition, and squares represent FAR condition.

Relationship between the CoM-CoP displacement and the trunk angle

Across subjects and conditions, there was no significant correlation between CoM-CoP displacement and trunk angle (r = -0.19, p = 0.177, Figure 9). Within each condition, trunk angle was positively associated with CoM-CoP displacement at maximum CoM-CoP displacement during swing for NO, NEAR, and FAR (r = 0.669, p = 0.003; r = 0.564, p = 0.018; r = 0.702, p = 0.002). Specifically, for each obstacle condition, increased CoM-CoP displacement is associated with increased trunk flexion at maximum CoM-CoP displacement during swing (Figure 9).

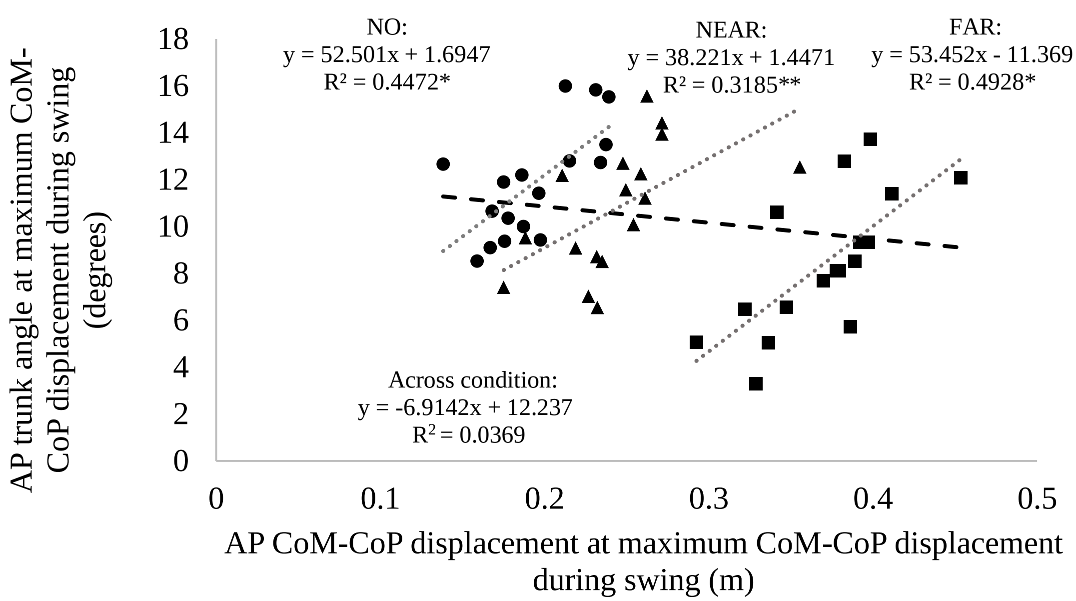

Figure 9. Relationships between AP CoM-CoP displacement and trunk angle at maximum CoM-CoP displacement during swing. Within each conditions is depicted by dotted regression lines, and across the conditions for individual subjects is depicted by dashed regression line in black color. Circles represent NO, triangles represent NEAR, and squares represent FAR condition. * significant relationship ($p<0.01$). ** significant relationship ($p<0.05$)

Within participants, we saw more significant correlations between CoM-CoP displacement and trunk angle at maximum CoM-CoP displacement during swing for the obstacle conditions (14/17 for both NEAR/FAR) than the no obstacle condition (10/17).

Variability of the CoM-CoP displacement and the trunk angle

With the increasing difficulty of the task, SD did not differ significantly for CoM-CoP displacement at maximum CoM-CoP displacement during swing between all the conditions ($p > 0.05$) whereas it differed significantly for trunk angle at maximum CoM-CoP displacement during swing between NO-NEAR, NO-FAR, and NEAR-FAR ($p = 0.044$, $p = 0.002$, and $p = 0.005$, Figure 10).

Within each condition, we observed no significant relationships between the SD of CoM-CoP displacement and trunk angle at maximum CoM-CoP displacement during swing ($r = 0.044$, $p = 0.867$; $r = 0.005$, $p = 0.985$; $r= 0.344$, $p = 0.177$ for NO, NEAR, and FAR, respectively, Figure 11). Similarly, across the conditions also, we observed no significant relationships between the SD of CoM-CoP displacement and trunk angle at maximum CoM-CoP displacement during swing for individual subjects' ($r = 0.089$, $p = 0.534$, Figure 11).

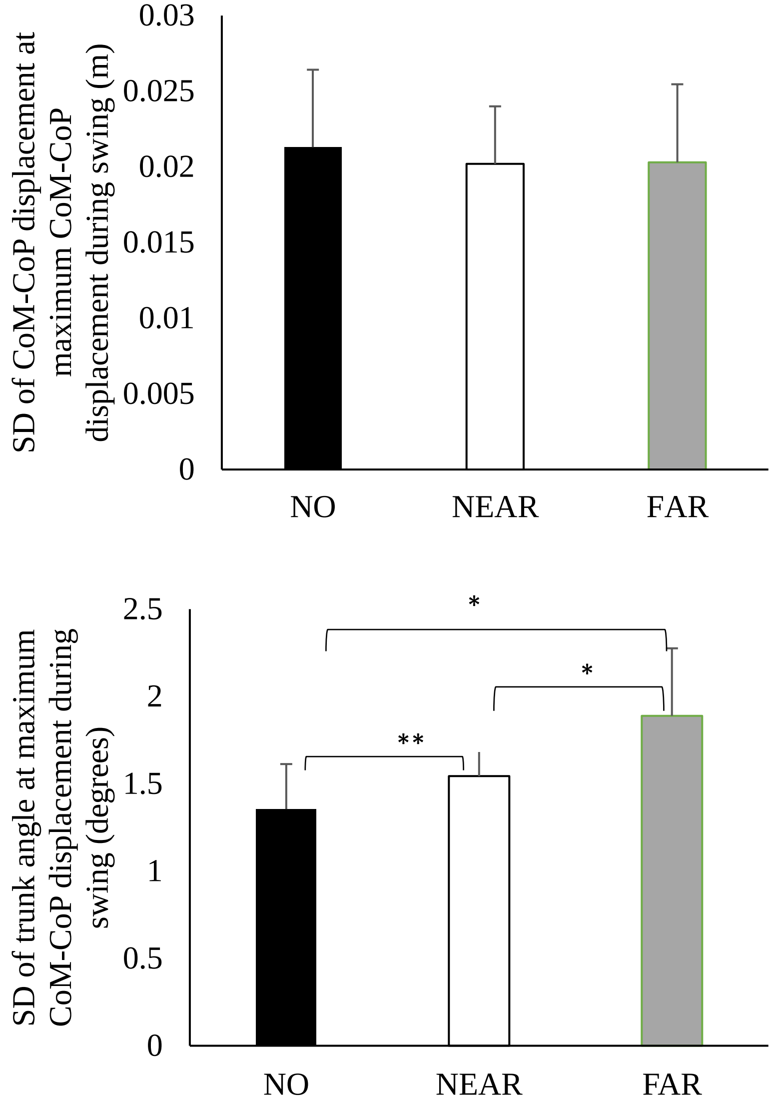

Figure 10. SD for CoM-CoP displacement and trunk angle at maximum CoM-CoP displacement during swing for all the conditions (mean±SD). *significant difference ($p<0.01$). **significant difference ($p<0.05$)

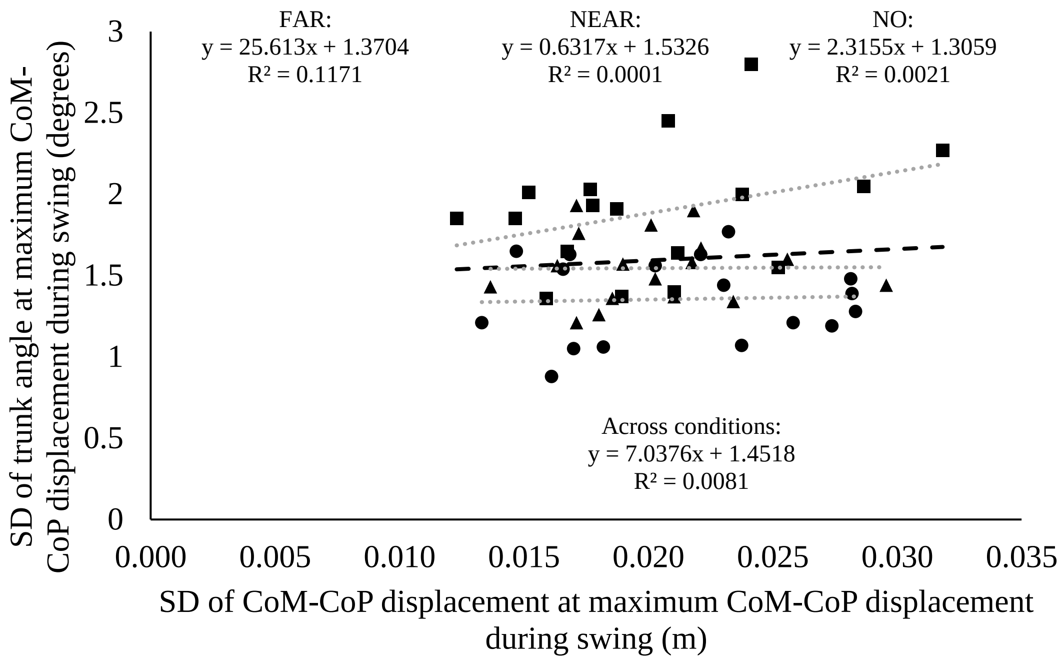

Figure 11. Correlation between SD for CoM-CoP displacement and trunk angle at maximum CoM-CoP displacement during swing. Within each condition is depicted by dotted regression lines and across the conditions for individual subjects is depicted by dashed regression line in black color. Circles represent NO, triangles represent NEAR, and squares represent FAR condition.

Discussion

The primary purpose of this study was to investigate postural control in a scaled GI task in the presence of spatio-temporal constraints imposed by an obstacle in healthy adults. There are three major findings in our study. At maximum CoM-CoP displacement during swing during GI: (1) CoM-CoP displacement increased significantly and trunk forward lean angle decreased significantly as the task became more challenging; (2) Variability of CoM-CoP displacement did not differ between the conditions whereas variability of trunk angle increased as the task became more challenging; and 3) Within each condition, CoM-

CoP displacement and trunk angle forward lean was positively correlated. Overall, our results suggest that healthy adults scale their postural control variables in the AP direction as the GI task becomes more challenging with near/far obstacles. This supports our hypothesis. Also, increased variability across conditions in the trunk angle may relate to the increased challenges of the task and decreased trunk control; however, no change in the variability of the CoM-CoP displacement across conditions may relate to the increased stability control by the postural control system.

Scaling of postural control variables in the AP direction

Maintaining postural stability is critical and can be achieved by modulating APAs or adjusting stability components in the swing phase during GI [61]. The scaling behavior of CoM-CoP displacement and trunk angle in the AP direction during the swing suggests that these variables might be sensitive measures of balance control for a scaled GI task constrained by an obstacle in healthy adults. AP CoM-CoP displacement during GI in healthy young adults is larger than in healthy older adults [27]. AP CoM-CoP displacement, while stepping over obstacles of increasing height, is reduced in young adults during the APA phase of GI [19]. However, in another study, increased AP CoM-CoP displacement during swing while stepping over obstacles of increasing height in young adults has been noted [41]. Moreover, Adkin et al. 2000 found that the magnitude of CoP displacement decreased with the increasing postural threat while standing quietly on a force plate raised to increasing heights from the ground [18]. It should be noted, though, that the challenge in GI is possibly a very different process than the challenge that fear of heights threat poses to quite standing tasks. From these studies, it seems that as the task becomes more challenging, postural control demand increases, and postural control variables decrease their amplitude potentially to maintain stability. However, with the scaled step distance task during GI in our study, we find that healthy adults respond to increasing postural challenges during the swing by increasing CoM-CoP displacement. The obstacle in our study acted as an additional task constraint causing healthy adults to take larger steps in the NEAR and FAR conditions with increased distance between the CoM and the CoP.

It was noteworthy that in response to the increasing step distance, the forward trunk angle decreased. Given that forward trunk lean is used to increase forward momentum [34] with increasing step distance, our results may at first be counterintuitive. Reduced trunk forward lean with the increasing postural challenge may serve to minimize the potential for instability that may arise from increased task difficulty. It may also reduce forward momentum of the step and help keep the CoM closer to the BOS during single-limb stance. Either could lead to instability during single limb support or at the initial step foot strike when initiating gait [12,72]. Indeed controlling trunk motions are important to maintain stability [73], and hence precise control of trunk forward lean becomes crucial. Moreover, it is also possible that a decrease in trunk forward lean with increasing difficultly of the task may be a function of body segments alignment resulting from taking a longer step to complete the task successfully. It is known that healthy individuals utilize the large mass of the trunk to affect whole-body movements [20]. Therefore, it can be suggested that healthy adults with good postural control keep their trunk upright by reducing their trunk forward lean to perform the task in a controlled/efficient manner.

Regulation between CoM-CoP displacement and trunk angle in the swing phase

As a part of our exploratory analysis, we found a positive relationship between CoM-CoP displacement and trunk angle at maximum CoM-CoP displacement during swing within each condition. This finding shows that healthy adults modulate CoM-CoP displacement and trunk angle to stabilize themselves during GI. In a previous finding, trunk kinematics during the swing phase and subsequent step width are correlated during gait [35], suggesting the possibility of a relationship between trunk motion in the swing phase and the planned step distance, which in turn could be related to CoM-CoP displacement to be achieved at foot strike. Therefore, there is a possibility that increased step distance task constraint necessitates an increase in CoM-CoP displacement for successful completion of the task, which may be related to increased trunk forward lean. With increased required step distance, increased trunk forward lean can be used to propel a body forward to bring CoM away from CoP under the stance limb basically to increase CoM-CoP displacement. There is some evidence that limiting trunk motion reduces step width during gait to maintain stability [27], which could be implied to our study findings where changing step distance, which primarily dictates CoM-CoP displacement might be further dictating trunk motion during GI. Additionally, given the large mass of trunk and the task constraints to extend the BOS, the ability of the trunk muscles to precisely control the trunk is limited. This could lead to passive trunk forward lean towards gravity. Thus, all of these factors could explain the relationship between AP CoM-CoP displacement and trunk lean at maximum CoM-CoP displacement during swing.

Variability in the postural control variables

Variability is considered a marker of neuromuscular function [75] and is considered an indicator of postural control during GI [76]. Findings from our study illustrate that variability in CoM-CoP displacement does not change with increasing postural challenge as CNS needs to have precise postural control to allow safe progression. However, trunk angle variability increases, potentially because of the involvement of the vestibular component. The vestibular system is aligned such that the canals and otoliths are perfectly horizontal when our head is forward by about 30 degrees [77]. As widely known, the strong interaction between trunk and head, it could be implied that the vestibular system that perceives as vertical is a slight lean forward. With the more difficult tasks in NEAR and FAR, that creates the need to increase the CoM-CoP displacement, which requires the trunk musculature to work harder to maintain the trunk upright. Therefore, one way to relieve that additional constraint is to align the trunk closer to vertical to reduce the gravitational forces acting on the trunk. Thus, the resulting increased trunk variability is a consequence of this "fight" between what the vestibular system perceives as "up" and reduction of gravitational forces on the trunk as it moves closer to vertical. This provides us an understanding of underlying sensorimotor control playing a role at different body segments [78]. It is apparent that when task challenges increase, there is a need for tighter control to avoid instabilities. Feedforward control mainly decides CoM-CoP displacement to be targeted to progress safely for task completion. This could lead to unchanged variability in CoM-CoP displacement, whereas, increased variability in the trunk angle might reflect instability created by task constraints. Moreover, the CNS is struggling to maintain stability with the increased task difficulty. Larger trunk variability reflects that struggle. Hypothetically if task difficulty continues to increase, without a concomitant increase in trunk variability, it can lead to loss of balance altogether. Correlation between variability of AP CoM-CoP displacement and trunk angle was not found significant within and across the conditions, which may indicate that variability at upper and lower body segments occur relatively independently during GI tasks.

Limitations

Small sample size is one of the limitations of our study. We had to exclude two participants from the analysis due to the inability to determine the pelvis markers location in one subject, and the other subject was making a toe strike instead of heel strike during trials, which made identification of the first step incalculable. With more participants, we may have found a significant relationship between the variability of CoM-CoP displacement and the trunk angle at maximum CoM-CoP displacement during swing within the conditions. The standard non-adjusted height of the obstacle could have affected our findings; however, obstacle height was within 18-22% of the individual's leg length for all the participants. Therefore, we do not expect this had much impact on our results. We asked all participants to start with their right leg [79,80] to make sure all the participants used the right leg for stepping, unlike other studies on stepping over obstacles where their participants used their preferred leg [19,81]. Notably, leg dominance does not affect postural stability in young adults [82]. We, therefore, do not expect differences in our results due to leg usage when performing the task. Even accounting for these minor limitations, our study findings will provide a reference to assess deviations/deficits in the balance parameters for clinical populations.

Conclusion

Our findings provide valuable information on trunk forward lean and CoM-CoP displacement during the swing phase of GI. Scaling of these postural control variables and their relationship in the presence of an obstacle reflects the regulation of balance under the increasingly destabilizing conditions. It can be emphasized that the increased CoM-CoP displacement is active due to the task constraints, and reduction of the trunk angle towards vertical is active to reduce gravitational forces acting on the trunk. When healthy adults transition from unperturbed upright stance to walking under progressively more challenging conditions due to obstacles at a varying distance, the postural control system regulates dynamic balance control to ensure that postural stability is regained at foot strike by maintaining the variability of CoM-CoP displacement. Whereas increased variability in the trunk angle is a consequence of the fight, not an active increase. It reflects the balancing act between the perception of vertical and reduction of trunk gravitational forces. Future studies on balance control during progressively challenging conditions during GI may also provide insight on balance and mobility in clinical populations with diminished balance control.

CHAPTER 3

EFFECT OF SCALED STEP DISTANCE TASK ON POSTURAL CONTROL VARIABLES DURING GAIT INITIATION IN OLDER NON-FALLERS AND FALLERS

Introduction

One in every four older adults aged ≥ 65 years fall each year in the US, and with increasing age, fall incidence in older adults increases as well [2]. Two common causes of falls by older adults are related to transfers and trips [8,9]. An insufficient transfer or shift in body weight, especially during gait initiation (GI), can result in a loss of control of the center of mass (CoM), resulting in a fall. Trips could result from improper control of the swing limb while advancing the base of support (BOS) during gait and thus can result in a fall. Postural stability, a vital component of independent and safe mobility, is maintained by controlling CoM kinematics with respect to the BOS and ensuring sufficient toe-obstacle clearance by the swing limb, which otherwise could lead to loss of balance or a fall [1,83]. The study of dynamic postural control in aging individuals often relies on measures such as the distance between the CoM and the center of pressure (CoP) during functional movements such as a GI task [13,15,27,48]. The distance between CoM and CoP is termed as CoM-CoP displacement. During GI, it has been shown that older fallers reduce the backward CoP shift in the anticipatory postural adjustment (APA) phase compared to older non-fallers and limit forward movement of the CoM to increase stability [14].

Previous literature suggests that two postural strategies, altering foot placement (changing step length or step width) and altering trunk motion, are recruited to enhance postural stability by controlling CoM [20,84]. Biomechanical analysis of stepping over obstacles, a common daily task, is used to assess balance and gait function in older adults [49,50] and has shown to increase step time in older adults, potentially to ensure proper clearance of the obstacle by the swing limb compared to typical GI [51]. It is also noteworthy that in older fallers, static and dynamic balance measures are reduced compared to older non-fallers [5–7]. For example, during GI, fallers show reduced first step length compared to non-fallers [15]. Therefore, modification of step length may provide insight into postural stability. Tasks of daily activities such as stepping over obstacles or water puddle or a pet around the house require scalability to perform the task and ensure safety successfully. Scaling or scalability in this study is defined as increases in task difficulty such as increases of the obstacle distance that lead to significant differences in task performance such as amplitude of CoM-CoP displacements from one trial to another [19]. However, scaling of postural control variables while stepping over obstacles at increasing step lengths/distances during GI has not been investigated in older fallers and non-fallers. There is a clear need to fill this gap to enhance scientific knowledge and clinical practice. Moreover, postural stability behavior in older fallers and non-fallers during scaled GI tasks is important to understand the control of CoM kinematics.

Control of forward body momentum without losing balance requires a certain separation between CoM and CoP. It is well-established that CoM-CoP displacement is directly proportional to horizontal CoM acceleration [85]. Notably, CoM-CoP distance during GI is considered a sensitive marker of dynamic stability in balance impaired individuals [26]. Larger required CoM-CoP distance during GI could increase demands on active dynamic balance control to maintain upright posture [25,27]. A previous study found that with increasing obstacle height, CoM-CoP displacement at maximum CoM-CoP displacement during swing is increased in older adults [41]. Another study found that during GI, peak CoM-CoP displacement during the swing phase is reduced in older adults with balance impairment compared to healthy older adults [26]. Interestingly, during walking, CoM-CoP displacement was larger at heel strike in fallers compared to non-fallers [86]. Peak CoM-CoP inclination angle, which is also suggestive of dynamic stability, was found to be decreased in fallers compared to non-fallers during walking and grew larger with increasing obstacle height irrespective of the groups [87]. However, CoM-CoP displacement has not been investigated in older fallers and non-fallers during GI when challenged with increasing obstacle distances. Knowledge of this variable will provide a more in-depth understanding of postural stability control in older adults, especially in older fallers, in response to tasks of increasing challenges during GI.

Trunk motion and CoM velocity play a vital role in the maintenance of dynamic balance control [29,30,88,89]. Evidence suggests that older adults increase trunk motions in anterior-posterior (AP) and medial-lateral (ML) direction when navigating an obstacle course to control CoM and successfully complete the task [40]. During walking, older non-fallers adapt trunk rotations to different walking speeds, unlike older fallers [38]. However, another study suggested that average trunk flexion during gait was similar between fallers and non-fallers [39], illustrating mixed literature on trunk control in older adults. During

GI, it is noted that trunk forward lean decreases in older women compared to young women [32]. However, there is very limited data on trunk motions in older adults during scaled GI tasks, especially with stepping over an obstacle as a constraint. The above findings highlight the importance of examining trunk motions in increasing postural challenging tasks in older fallers and non-fallers. Overall, this may help to enhance our understanding of the role of trunk control in postural stability.

The AP velocity of the CoM at maximum CoM-CoP displacement during swing, where CoM-CoP separation is maximum during GI, also needs to be controlled to avoid instability. In older adults, AP CoM velocity at maximum CoM-CoP displacement during swing decreased linearly with increasing obstacle height during walking [41]. During walking, older fallers reduce peak AP CoM velocity compared to non-fallers [42]. Moreover, the AP velocity of the CoM at the heel strike during GI is lower in older adults than young adults [90]. All of the above findings make CoM velocity an important measure to be studied in older adults at greater risk for instability during scaled GI tasks. Moreover, CoM velocity has not been investigated in increasing challenging scaled GI tasks in older fallers and non-fallers.

Challenging postural stability by modifying stepping over obstacles accompanied by increasing distances in older adults, especially in balance impaired older adults, will stress the postural control system and help discriminate the postural control variables. In GI, AP and ML stability are required to progress forward and shift weight from one limb to the other. With increasing obstacle distance tasks during GI, it is suggested that young adults modulate APAs in the AP but not in the ML direction [19]. Therefore, focusing on investigating AP stability during AP scaled tasks would be important. In this study, we are focusing on postural control variables such as CoM-CoP displacement, CoM velocity, and trunk angle, which play a critical role in CoM control in older fallers versus non-fallers. The purpose of the study was to compare the scaling of the postural control variables between older non-fallers and fallers to a scaled step distance task during GI in the presence of spatio-temporal constraints imposed by an obstacle. We used the task of stepping over an obstacle during GI to determine the scaling behavior of older fallers and non-fallers. Primary variables of interest were the peak AP CoM-CoP displacement during APA, AP CoM-CoP displacement at the toe-off, and AP CoM-CoP displacement at maximum CoM-CoP displacement during swing. Secondary variables of interest were peak AP trunk angle during swing phase and CoM velocity at maximum CoM-CoP displacement during swing. We hypothesized that during GI, compared to non-fallers, older fallers will not scale their posture control, and that the fallers will not show scaling of 1) peak AP CoM-CoP displacement during APA and 2) AP CoM-CoP displacement at the toe-off and, 3) AP CoM-CoP displacement at maximum CoM-CoP displacement during swing while performing progressive stepping distances resulting in group x step distance interaction.

Materials and methods

Participants

We recruited 24 older adults (8 males, 16 females, age: 72.54 ± 4.77 years, height: 166 ± 12 cm, mass: 70.61 ± 11.02 kg) from the Birmingham, AL community. All participants provided written informed consent prior to their participation in this study approved by the Institutional Review Board of the University of Alabama at Birmingham

(IRB# 300002052). Participants had no known neurological, musculoskeletal, or cardiovascular conditions that could have affected their walking, and all were independent walkers. All the participants had an adequate cognitive function to participate as indicated by a score of more than 24 on the Mini-Mental State Exam and were categorized into two groups as older fallers and older non-fallers based on their history of at least one fall in the past 12 months. Based on this categorization, we had 12 participants in each group (non-fallers and fallers), as shown in Table 3. We also collected data on Berg Balance Scale, Timed-Up and Go test, and Activities-Specific Balance Confidence Scale scores for exploratory analysis. The World Health Organization definition of a fall (i.e., A fall is defined as an event which results in a person coming to rest inadvertently on the ground or floor or other lower-level") was provided to participants to assess their fall status.

Table 3. Descriptive characteristics (mean ± SD)

Variable	**Older non-fallers (n=12)**	**Older fallers (n=12)**
Age (yrs)	73.41 ± 3.91	71.66 ± 5.53
Gender (M/F)	5/7	3/9
Height (m)	1.66 ± 0.14	1.64 ± 0.098
Weight (kgs)	69.81 ± 8.77	71.40± 13.25

Protocol

Participants wore their shoes during the data collection. For all the trials, a spotter was present to avoid potential falls while performing the tasks. All participants were instructed to start walking at their preferred speed from a standing position with both feet on one force plate. Participants were asked to initiate gait with their right leg always. Instructions were: “when you hear ‘GO’ step forward with your right leg first, over the obstacle if there is one, and walk forward 3-4 steps and then return to your starting position. Perform all the trials at your comfortable speed, and you can look straight ahead or, if needed, look down to perform the trials”. The scaled stepping task included four experimental conditions. All participants were provided three practice trials for each condition. We calculated the normal standing step width for each participant by taking an average of the three step widths from their practice trials to standardize standing step width across each subject’s trials. The four experimental conditions were no obstacle (NO), near obstacle distance (NEAR), middle obstacle distance (MIDDLE), and far obstacle distance (FAR). Each condition had five trials, and participants completed 20 experimental trials in total. NO condition was defined as the absence of any obstacle in the participants’ walkway. NEAR, MIDDLE, and FAR conditions were defined as an obstacle placed at 25%, 45%, and 65% of the individual’s leg length, respectively. To mark obstacle distances on the walkway, we measured distances from an individual’s starting toe position on the force plate. Participants were able to see the walkway and obstacle (if placed per the conditions) throughout the trials. We had two blocks of trials where one of the blocks included five trials each of NO and NEAR, and the other block included five trials each of MIDDLE and FAR conditions. We chose these two blocks in an alternating sequence with successive participants. The trials within both blocks were randomized. We gave a rest period of ~10-15 minutes between the two blocks. A representative picture of a participant performing a trial of the experiment is shown in Figure 12.

Figure 12. Participant stepping for the NEAR obstacle condition on the 2nd force plate

Data collection and analysis

Vicon motion capture system using 11 infrared cameras (Vicon Motion Systems, Oxford, UK) and AMTI force plates (Advanced Mechanical Technology Inc., Newton, MA) were used to simultaneously collect kinematic and kinetic data, respectively. A 12-segment rigid body model was used to calculate kinematic data. Thirty-six reflective markers were placed over upper arms, lower arms, trunk, pelvis, upper legs, lower legs, and feet to collect kinematic data at 100 Hz. Upper arm segments had markers placed on the left/right acromion, left/right lateral epicondyle, and on the anterior aspect of the upper arm. Lower arm segments had markers placed on the left/right lateral epicondyle, posterior aspect of the wrist, and on the posterior aspect of the forearm. The trunk segment had markers placed on left/right acromion, left/right iliac crest, manubrium, and the 7th cervical vertebra. The pelvis segment had markers placed on the left/right anterior and posterior superior iliac spines. Upper leg segments had markers placed on left/right anterior superior iliac spines, medial/lateral femoral epicondyles, and on the anterior aspect of the thigh. Lower leg segments had markers placed on medial/lateral epicondyles, medial/lateral malleoli, and the anterior aspect of the shank. Feet segments had markers placed on medial/lateral malleoli, first/fifth metatarsophalangeal joints, and on the heel. Two AMTI force plates placed successively in forward direction were used to collect kinetic data at 1000 Hz. To collect, process, and analyze biomechanical data, we used Vicon Nexus, Visual 3D (C-Motion Inc., Rockville, MD, USA), and MATLAB 2018a (MathWorks Inc., Natick, MA, USA) software.

Kinematic and kinetic data were filtered with a 4th order low-pass Butterworth filter with a cutoff frequency of 6 Hz for kinematic and 12 Hz for kinetic data. Three-dimensional spatial coordinates of the markers placed on the body were used to calculate whole-body CoM, CoM velocity, and trunk angle using Visual 3D software (C-Motion, Inc., Rockville, MD, USA) and; ground reaction force data was used to calculate whole-body CoP, toe-off of the right foot, and right foot-strike using Visual 3D software. Custom MATLAB 2018a software was used for further calculations. CoM-CoP displacement was calculated as the distance between CoM and CoP position in the AP direction. Step length was calculated using the right heel data from static standing to the right heel strike of each trial. Step time was calculated from right foot toe-off to right foot strike. For each trial, we extracted data of the 1st step. Data at the time of immediately prior to the right foot strike, when the distance between CoM and CoP reached maximum, were extracted for analysis.

AP CoM-CoP displacement data was extracted for the peak value during APA, at the toe-off, and at maximum CoM-CoP displacement during swing for each trial. AP trunk angle data was extracted for the peak value during swing for each trial. AP CoM velocity data was extracted at maximum CoM-CoP displacement during swing for each trial. Data were then averaged across trials in each condition to calculate the mean and standard deviation of all variables of interest.

Statistical analysis

We used within and between repeated measures ANOVA (rmANOVA) to compare means between groups for each condition for an interaction effect (Group x Condition) and main effects (Group and Condition) for peak AP CoM-CoP displacement during APA, AP CoM-CoP displacement at the toe-off, AP CoM-CoP displacement at maximum CoM-CoP displacement during swing, peak AP trunk angle during swing, CoM velocity at maximum CoM-CoP displacement during swing. For descriptive purposes, we ran rmANOVA on spatio-temporal variables, which were step length and step time (2 groups x 4 conditions). All of the statistical analyses were conducted using SPSS software (version 25.0; IBM Corporation, Armonk, NY), and alpha was set at 0.05 for all the analyses. Bonferroni post-hoc adjustments were used to adjust for multiple comparisons.

Results

Effect of task on step length and step time

For step length, there was no main effect of group ($p > 0.05$); however, there was a main effect of condition ($p < 0.001$). Participants took longer steps from NO to NEAR, NO to MIDDLE, NO to FAR, NEAR to MIDDLE, MIDDLE to FAR, and NEAR to FAR (p <

0.001 for all, Table 2). We also found no significant interaction between group and condition ($p > 0.05$).

For step time, we found a significant main effect of condition and group ($p < 0.001$ for both). However, we also found a significant interaction between group and condition ($p < 0.001$). For condition NO, fallers and non-fallers had no difference in the step time ($p = 0.884$), however, fallers had longer step time in NEAR, MIDDLE, and FAR than non-fallers ($p = 0.035$, $p = 0.001$, and $p < 0.001$, respectively, Table 4).

Table 4. Step length and step time for all the conditions (mean ± SD) for both the groups

	Non-fallers		**Fallers**	
Obstacle Condition	Step length (m)	Step time (s)	Step length (m)	Step time (s)
NO	0.54 ± 0.07	0.41 ± 0.04	0.53 ± 0.09	0.41 ± 0.04
NEAR	0.66 ± 0.06	0.71 ± 0.06	0.64 ± 0.06	0.79 ± 0.11
MIDDLE	0.81 ±0 .05	0.72 ± .06	0.79 ±0 .07	0.84 ± 0.09
FAR	0.93 ± 0.06	0.74 ± 0.07	0.93 ± 0.06	0.88± 0.10

Effect of task on CoM-CoP displacement

For peak AP CoM-CoP displacement during APA, we noted a significant main effect of group ($p = 0.018$) and condition ($p < 0.001$, Figure 13). However, no significant group x condition interaction ($p > 0.05$) was found. Fallers had smaller peak CoM-CoP displacement than non-fallers irrespective of conditions. Peak AP CoM-CoP displacement during APA was smaller in NEAR than NO ($p < 0.001$), in MIDDLE than NO ($p < 0.001$), in NEAR than MIDDLE ($p = 0.030$), in NEAR than FAR ($p = 0.001$), and in MIDDLE than FAR ($p = 0.002$). An example of individual data points of peak AP CoM-CoP displacement for the FAR condition is shown in Figure 14.

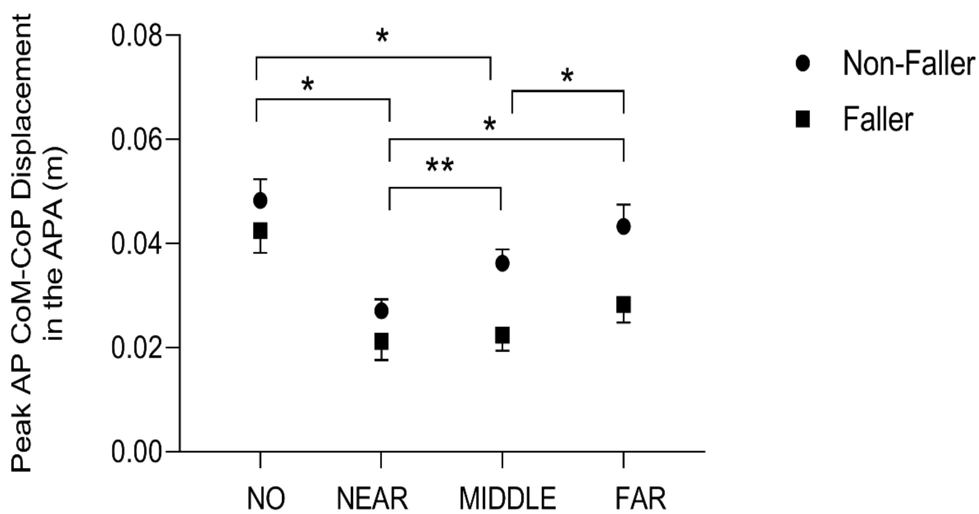

Figure 13. Peak AP CoM-CoP displacement during APA for all the four conditions. * represents p<0.01. ** represents p<0.05.

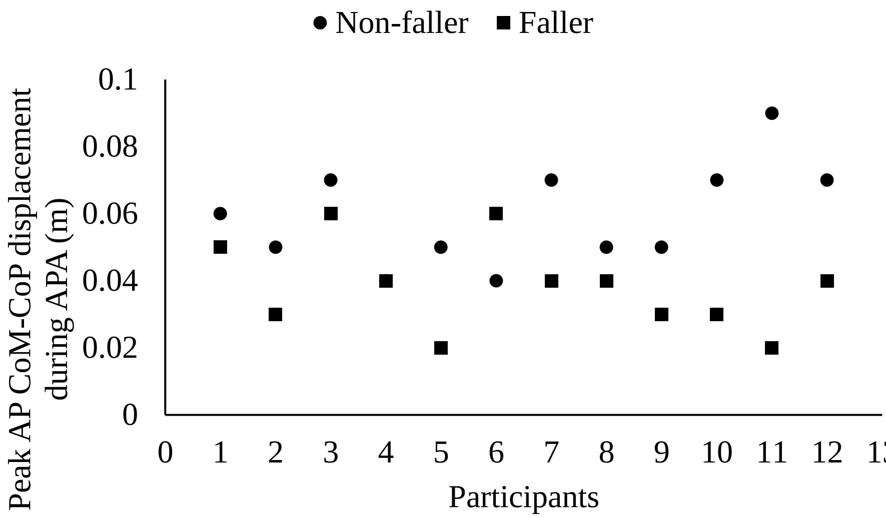

Figure 14. Data of individual participants for peak AP CoM-CoP displacement during APA performing the FAR condition. Circles represent participants in the non-fallers group and squares represent participants in the faller group.

For AP CoM-CoP displacement at the toe-off, we observed a significant main effect of group (p = 0.016) and condition (p < 0.001). We also found a significant group x condition interaction (p = 0.002, Figure 15). For condition NO & NEAR, fallers and non-fallers had no difference in AP CoM-CoP displacement at the toe-off (p = 0.439); however, for condition MIDDLE and FAR, fallers had smaller AP CoM-CoP displacement at the toe-off than non-fallers (p = 0.013 and p = 0.003, respectively).

For AP CoM-CoP displacement at maximum CoM-CoP displacement during swing, we found a significant main effect of condition (p < 0.001) but no significant main effect of group (p > 0.05, Figure 16). There was no significant group x condition interaction (p >0.05). AP CoM-CoP displacement increased at maximum CoM-CoP displacement during swing from condition NO to NEAR, NO to MIDDLE, NO to FAR, NEAR to MIDDLE, MIDDLE to FAR, and NEAR to FAR ($p < 0.001$ for all).

Effect of task on trunk angle and CoM velocity

For the peak AP trunk angle during the swing phase, we found a significant main effect of group ($p = 0.014$) and condition ($p < 0.001$, Figure 17). We observed no significant group x condition interaction ($p > 0.05$). Fallers had higher peak trunk forward lean during swing than non-fallers. Peak trunk flexion was larger from NO to NEAR, NO to MIDDLE, NO to FAR, NEAR to FAR, and MIDDLE to FAR ($p = 0.002$, $p = 0.011$, $p < 0.001$, $p = 0.001$, and $p < 0.001$, respectively). An example of individual data points of peak AP trunk angle for FAR condition is shown in Figure 18.

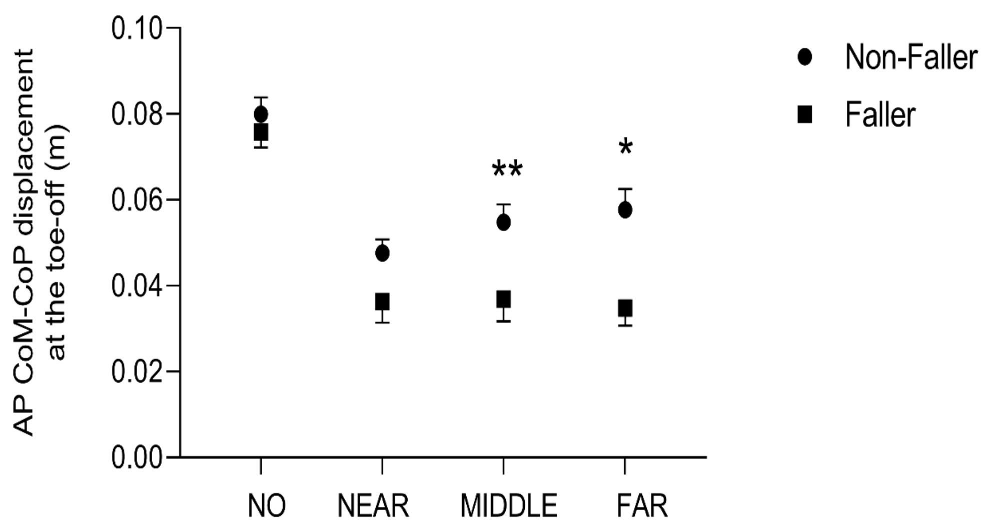

Figure 15. AP CoM-CoP displacement at the toe-off for all the four conditions. * represents p<0.01. ** represents p<0.05.

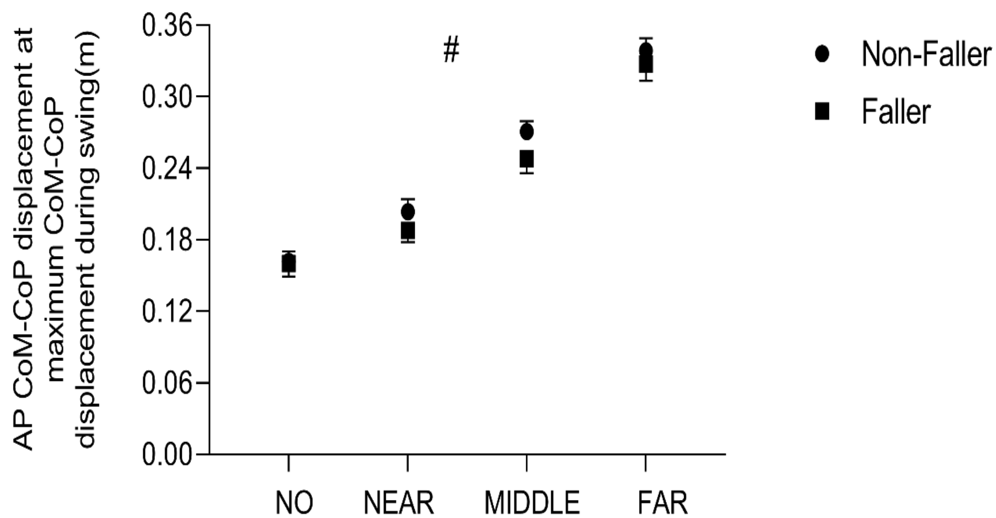

Figure 16. AP CoM-CoP displacement at maximum CoM-CoP displacement during swing for all the four conditions. # represents all the conditions were significantly different from each other where p<0.05.

For AP CoM velocity at maximum CoM-CoP displacement during swing, we found a significant main effect of condition ($p < 0.001$), but no significant main effect of group ($p > 0.05$, Figure 19). There was no significant group x condition interaction ($p > 0.05$). Participants increased CoM velocity at maximum CoM-CoP displacement during swing from condition NO to NEAR, NO to MIDDLE, NO to FAR, NEAR to MIDDLE, MIDDLE to FAR, and NEAR to FAR ($p < 0.001$ for all).

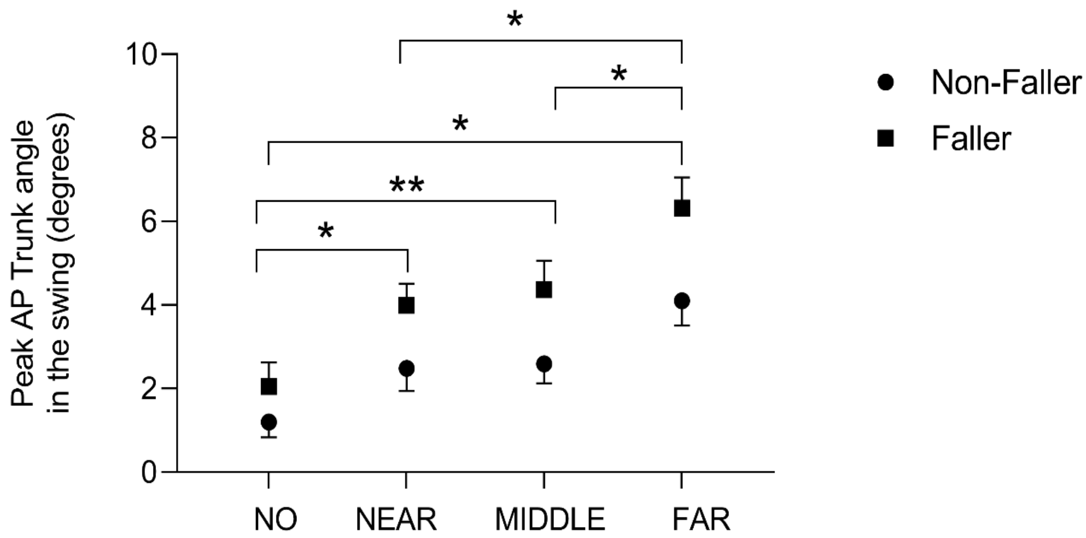

Figure 17. Peak AP trunk angle during swing for all the 4 conditions. *indicates statistical significance for p<0.05

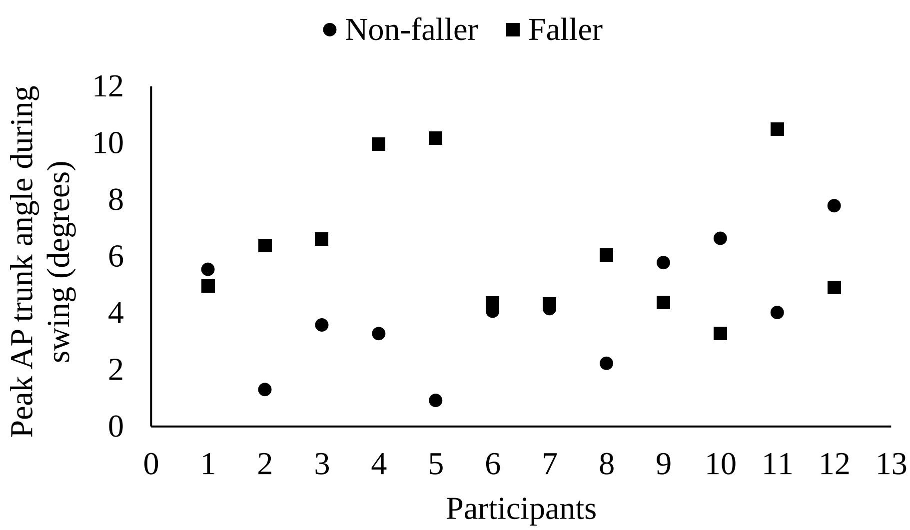

Figure 18. Data of individual participants for peak AP trunk angle during swing performing the FAR condition. Circles represent participants in the non-fallers group and squares represent participants in the faller group.

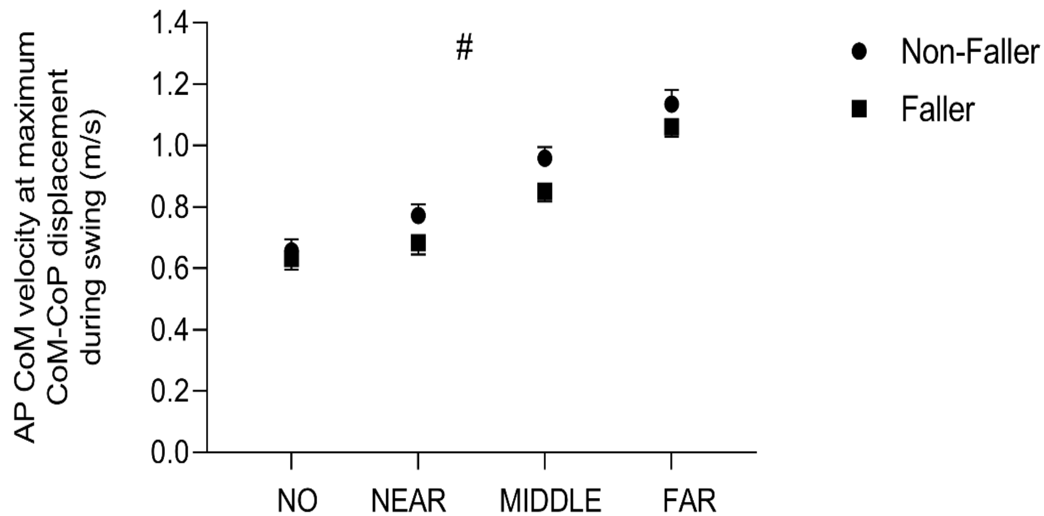

Figure 19. AP CoM velocity at maximum CoM-CoP displacement during swing for all the four conditions. # represents all the conditions were significantly different from each other where $p<0.01$.

Discussion

Our study investigated differences in postural control variables between older fallers and non-fallers while stepping over obstacles at progressively increasing distances during GI. In general, our results demonstrate that older fallers display reduced CoM-CoP displacement prior to the toe-off and an increased trunk lean during swing for more challenging conditions compared to non-fallers. The major findings are 1) older fallers had smaller peak AP CoM-CoP displacement in the APA phase compared to non-fallers; 2) at the toe-off, older fallers reduced AP CoM-CoP displacement in MIDDLE and FAR conditions compared to non-fallers but not in NO and NEAR conditions; and 3) peak forward trunk lean during swing was larger in older fallers compared to non-fallers. Largely, these findings imply that older fallers modify postural control variables to potentially minimize increased task-related postural demand to enhance their stability for successful task completion.

AP CoM-CoP displacement behavior

AP CoM-CoP displacement in our study could imply that it may be a sensitive marker of postural stability. With increased AP CoM-CoP displacement, disequilibrium torque generated to propel the body forward increases [91]. It can potentially increase the chances of instability at toe-off due to increased CoM acceleration (CoM-CoP displacement is directly proportional to CoM acceleration) during GI. With increased AP CoM-CoP displacement, an increased active postural control is also required. Specifically, in the APA phase, an increased AP CoM-CoP displacement could lead to an increased CoM forward acceleration [13]. This could further destabilize individuals with decreased postural control ability and may lead to instability or uncontrolled fall. Fallers have been observed to reduce AP CoP shifts during APA in challenging conditions compared to non-fallers [92]. Our reduced AP CoM-CoP displacement during APA in fallers is in line with this previous study. The most common cause of falls is during shifting of bodyweight that causes the CoM to move beyond the BOS (or improper transfers) in older adults [8]. In the transition phase from double limb support to single-limb support, fallers try to keep AP CoM-CoP displacement at the toe-off smaller than non-fallers. This could occur because, at toe-off, there are increased risks of losing balance control and instability as one transfers weight from one limb to the other. Therefore, fallers, due to reduced postural control ability

[93], may try to keep AP CoM-CoP displacement smaller compared to non-fallers in the APA and at the toe-off. However, at maximum CoM-CoP displacement during swing, both fallers and non-fallers had to increase AP CoM-CoP displacement because of the task demand. Both fallers and non-fallers successfully completed the task by scaling AP CoM-CoP displacement at maximum CoM-CoP displacement during swing similarly. A similar degree of task-requirement specific increases, in fallers and non-fallers, for AP CoM-CoP displacement at maximum CoM-CoP displacement during swing reflects on prioritization of task completion over increased postural challenges posed by the task. Overall, AP CoM-CoP displacement findings suggest that when challenged with an increased level of difficulty, fallers reduce AP CoM-CoP displacement before the toe-off but could not reduce AP CoM-CoP displacement before the initial foot contact. It seems that the modulation of AP CoM-CoP displacement in response to increased postural control demand is GI-events-specific in fallers.

Trunk motion, CoM velocity and step time

In our findings, with increased task challenges, there was an increased forward trunk flexion in older non-fallers and fallers. It could be that older non-fallers and fallers specifically did this to ensure they could complete the task. Reduced AP CoM-CoP displacement at maximum CoM-CoP displacement during swing reduces forward-directed velocity, and thus to achieve the AP CoM velocities as seen in Figure 5, they needed to lean the trunk farther.

Trunk flexion was larger in fallers compared to non-fallers. This may be due, at least partially, to the findings that fallers are known to have reduced gluteus maximus activity and, thus, reduced rate of hip torque compared to non-fallers [94]. Therefore, fallers may pre-activate gluteus maximus and hip extensors to allow better muscle force production for successful task completion. It should be noted that an increased trunk forward flexion could assist in taking swing limb forward while clearing the obstacle. In the swing phase, an increased forward trunk lean in fallers can also bring CoM forward toward the stepping limb. Since CoP is closer to the stance limb, a certain distance between CoM and CoP can thus be created to help propel the body forward and thus could compensate for weaker hip muscles.

Muscle activity also gets reduced in fallers [95]. Fallers, thus, cannot meet the demand for increased muscle strength required to maintain trunk and head upright against gravity that leads to an increased trunk forward lean. Moreover, with aging, there is an increased contribution of hip muscles to ankle muscles in gait [96]. Therefore, in fallers, weaker hip muscles, and reduced muscle activity could be unable to meet the demands of an increased contribution of the hip, potentially resulting in greater trunk flexion. It has been seen in a previous study that trunk control (increased trunk flexion angle) at the initial contact is related to keeping the CoM closer to the BOS for a safe landing, as seen in soccer players during landing [97]. However, in our study with the GI task, an increased peak trunk angle during swing could be to control the CoM movement in a way to help accelerate the body forward.

In our study, step time was increased in fallers compared to non-fallers, perhaps as a cautious behavior to execute the step. Further, our exploratory analysis showed a positive correlation between step time and peak AP CoM-CoP during APA in fallers but not in non-fallers. Previous research has suggested that young adults increase step time when crossing obstacles during rapid triggered stepping reactions, and it was also found that there was an increased APA in ML direction [68]. Similar to their finding, in our study, fallers with increased step distance increased step time and utilizing APAs in the AP direction to potentially increase stability. There is also a possibility that fallers by increasing step time are showing poor postural control ability and therefore displaying a cautious behavior although paradoxically, may increase the risk of falls.

Finding the increased AP CoM velocity at maximum CoM-CoP displacement during swing per the challenging tasks in our study indicates faster completion of the task hypothetically to reduce the time spent on single-limb support in older adults. However, no difference in the AP CoM velocity at maximum CoM-CoP displacement during swing between fallers and non-fallers could suggest a non-optimal scaling of the AP CoM velocity in fallers. This may be explained by increased step time in fallers than non-fallers. With an increased step time and a comparable AP CoM velocity at maximum CoM-CoP displacement during swing, fallers show a cautious approach.

Limitations

A limitation of the study is that we always asked all participants to start the task using their right leg, which could have affected our findings. Differences in leg dominance between participants could impact how they perform the task and could have affected our results. However, our method of using the right leg always is in line with a previous study, which also used the right leg for all of their GI or obstacle avoidance during GI tasks [51]. Using the right leg in our study allows us an opportunity to compare our data with other studies. Even though we normalized obstacle distance with participants' leg length, fixed obstacle height could have affected how fallers and non-fallers cleared the obstacle, affecting variables of our study. However, no significant differences in the height of participants between the groups indicate that it would not have affected our results. All participants started from their average stance width, which could have affected how participants performed the tasks. Standardizing stance width in our experimental setup could have changed our results, although it would not have been reasonable to ask participants to start from their uncomfortable stance width.

Conclusion

Older fallers demonstrated smaller CoM-CoP displacements suggestive of a more conservative strategy than older non-fallers prior to the toe-off while performing challenging AP tasks constrained by obstacles during GI. However, during the step, older fallers used a larger forward trunk lean, which may suggest that to complete the task successfully, fallers utilize trunk motions. It can also be suggested that differences in postural stability between fallers and non-fallers are most evident with CoM-CoP displacement during APA and with the trunk angle during swing phase of GI while performing tasks.

CHAPTER 4

COMPARATIVE SCALING OF POSTURAL CONTROL VARIABLES IN SCALED STEP WIDTH TASKS DURING GAIT INITIATION IN OLDER NON-FALLERS AND FALLERS

Introduction

Postural instability is a chief factor associated with falling in older adults [98]. Falls due to impaired postural stability in the medial-lateral (ML) plane is common in older adults, resulting in laterally-directed falls [8,98,99], and can lead to injuries (i.e., hip fracture), which are associated with significant mortality [100]. Control of ML stability is crucial to avoiding lateral falls. During gait initiation (GI), active postural control of ML stability is required to transition from a quiet stance to steady-state walking [101]. Notably, the GI task has been used to investigate impairments in postural stability [102,103]. During GI, ML anticipatory postural adjustment (APA) and swing limb foot placement play an important role in ML stability control [68,104–106].

Foot placement at the end of the swing phase of GI creates a new base of support (BOS), and if the center of mass (CoM) falls outside the BOS, postural instability may occur. During GI tasks, when individuals walk at different speeds, young adults take wider steps to compensate for an increased CoM shift laterally and thus to maintain ML stability [102]. Importantly, lateral foot placement has been considered as one of the primary mechanisms for ML stability control in younger adults [56]. Interestingly, the effect of a wider stride on postural stability is unclear in older adults [98]. Movement tasks that force a narrow step width are common ways to challenge ML stability in older adults [21,24,107]. Older adults decrease the margin of stability when walking with a narrow base than young adults [24]. In older fallers, ML postural sway increases with a narrow base stance [5]. Moreover, targeting different step widths like wide and narrow could affect ML stability during walking in older adults [11,21,108]. Notably, this has not been studied in older adults, especially in fallers during GI, which is of clinical significance to understand strategies adopted by fallers to maintain ML stability. In GI, anterior-posterior (AP) and ML stability are required to progress forward and shift weight from one limb to the other. It is suggested that during rapid triggered stepping reaction tasks, young adults utilize lateral foot placement to control ML stability [106]. Similarly, it can be postulated that modifying ML foot placement would stress the frontal/ML stability control in older adults. In theory, it would require modification to their APA and the swing phase, providing a means to investigate differences in postural stability between older non-fallers and fallers.

Obstacle clearance affects the swing phase by increasing swing duration to control ML stability during triggered forward stepping in young adults [68,106]. With the increase in swing duration because of obstacle presence, there is an increased lateral movement of the center of pressure (CoP) during the APA phase and lateral foot placement at the heel strike [106]. There is evidence that older adults with balance impairment compared to healthy older adults increase ML CoM motion and ML swing limb excursion when stepping over obstacles during walking to maintain ML stability [109]. It is also reported that, older adults increase trunk motions during such a task but adopt a narrow step width [110]. All of these demonstrate that limited knowledge is available on the modulation of

ML APA and swing phase in older non-fallers and fallers, especially when targeting different step widths during GI while stepping over an obstacle. This will help broaden the understanding of postural stability during ML postural challenging tasks in older adults.

One of the important measures of postural stability is CoM-CoP displacement during the movement task. CoM-CoP displacement in older adults with balance impairment is considered a sensitive marker of stability [67]. In GI, peak ML CoM-CoP distance during the end of the swing limb foot strike was found to be smaller in more versus less balance impaired individuals [27]. However, the ML CoM-CoP displacement at the heel strike is not different in non-fallers versus fallers during level walking [52]. Interestingly, previous research has suggested that the ML CoP movement during GI was not different in fallers and non-fallers [14]. However, requiring different step widths during GI may affect their ML APA and swing phase and is thus important to be investigated to provide insight into different postural stability control between fallers and non-fallers.

The trunk is a large mass, and trunk control has been suggested to be an important variable while initiating gait [71], walking [35], and recovery from a postural disturbance [36]. The trunk range of motion in the transverse plane has been reported to not differ between older fallers and non-fallers when walking speed was changed [38]. Older adults increase the trunk range in the frontal plane when navigating through an obstacle course during walking to control CoM [40]. A recent article suggested that constraining trunk movements affect step width during walking, implying a coupling between trunk motion and foot placement [74]. Such results suggest that trunk angle kinematics, which plays an important role in CoM positioning relative to the base of support (BOS), maybe a crucial variable to be studied in older adults to gain knowledge on trunk regulation when performing tasks of different step width during GI.

Another important measure to maintain ML stability is ML CoM velocity, which, if increased near the foot strike, could require an increased hip abductor moment or a compensatory step to avoid instability [11]. During GI, the ML velocity of the CoM at initial contact in older adults is lower than that of young adults, probably to compensate for increased CoM lateral movement in older adults [90]. Interestingly, peak ML velocity of the CoM while stepping over obstacles at different heights during walking is greater in older fallers than in non-fallers [109]. Moreover, information on ML CoM velocity while performing wider or narrower step widths during GI could help understand postural stability in older fallers and non-fallers.

To the best of our knowledge, no information exists on the above-mentioned postural control variables (CoM-CoP displacement, trunk angle, CoM velocity) in older fallers versus non-fallers while performing obstacle clearance tasks with medial and lateral scaled step widths during GI. Therefore, the purpose of the study was to determine the extent to which older fallers and non-fallers scale the postural control variables when stepping over an obstacle while increasing or decreasing step width to challenge ML stability. We used the task of stepping over an obstacle with increased and decreased step widths during GI to determine the scaling behavior of older fallers and non-fallers. Scaling or scalability in this study is defined as changes in task difficulty such as scaled step widths that lead to significant differences in task performance such as amplitude of CoM-CoP displacements from one trial to another [19]. Primary variables of interest were the peak

ML CoM-CoP displacement during APA, ML CoM-CoP displacement at the toe-off, and ML CoM-CoP displacement at maximum CoM-CoP displacement during swing. Secondary variables of interest were peak ML trunk angle during swing and ML CoM velocity at maximum CoM-CoP displacement during swing. As illustrated in Figure 20, we hypothesized that older fallers performing a GI task with targeted initial step widths would result in no scaling of A) peak ML CoM-CoP displacement during APA and B) ML CoM-CoP displacement at the toe-off and, C) ML CoM-CoP displacement at maximum CoM-CoP displacement during swing compared to non-fallers resulting in group x step distance x step width interaction.

Materials and methods

Participants

We recruited 24 older adults (5 males, 15 females; age: 72.54 ± 4.77 years; height: 166 ± 12 cm; mass: 70.61 ± 11.02 kg) from Birmingham, AL community. We obtained written informed consent from all the participants prior to their participation in this study approved by the Institutional Review Board of the University of Alabama at Birmingham (IRB# 300002052). All participants were independent walkers and had no known neurological, musculoskeletal, or cardiovascular conditions that could have affected their walking. To assess the cognitive function of participants, we conducted the Mini-Mental State Exam in which participants had to score >24 to participate in the study. We also collected data on Berg Balance Scale, Timed-Up and Go test, and Activities- Specific Balance Confidence Scale scores for exploratory analysis. The World Health Organization definition of a fall (i.e., “A fall is defined as an event which results in a person coming to

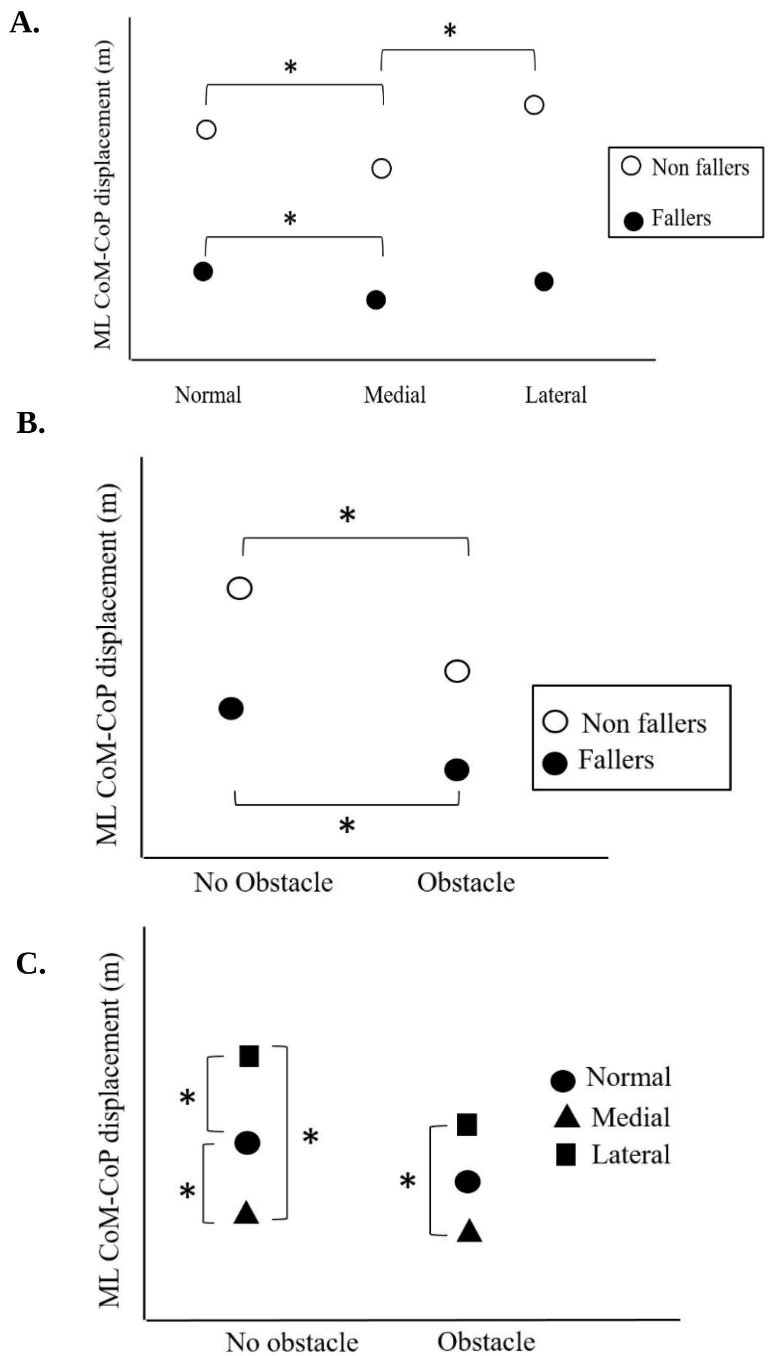

Figure 20. Depiction of hypothesis. Older fallers performing a GI task with targeted initial step widths would result in no scaling of A) peak ML CoM-CoP displacement during APA, B) ML CoM-CoP displacement at the toe-off, and C) ML CoM-CoP displacement at maximum CoM-CoP displacement during swing compared to non-fallers resulting in group x step distance x step width interaction.

rest inadvertently on the ground or floor or other lower-level") was provided to participants to assess their fall status. Participants were categorized based on their history of at least one fall in the past 12 months into two groups as older fallers and older non-fallers. We had 12 participants each in older non-fallers and fallers group. Due to unforeseen limitations in the size of our force plates and the preferred stance width of participants, we were unable to collect complete data on 4 participants. Coincidently, all these 4 participants were fallers. Therefore, we had 8 participants in the fallers group. The descriptive data of participants for each group is shown in Table 5.

Table 5. Participants' physical characteristics (mean ± SD)

Variable	**Older non-fallers (n=12)**	**Older fallers (n=8)**
Age (yrs)	73.41 ± 3.91	70.62 ± 5.26
Gender (M/F)	5/7	0/8
Height (m)	1.66 ± 0.14	1.59 ± 0.057
Weight (kgs)	69.81 ± 8.77	70.23 ± 12.69

Protocol

We asked our participants to wear their shoes during the data collection. A spotter stood close, but behind participants to assist in the event of a loss of balance while stepping for all the trials. Participants were asked to start from a fixed standing position with both feet on one force plate and always initiate gait with their right leg. Instructions were: "when you hear 'GO' step over the obstacle using your right leg first if you see the obstacle in your walkway or start walking normally using your right leg first if you see no obstacle in your walkway and using the same right leg step on the specific target lines if mentioned otherwise step normally and then continue to walk for 3-4 steps normally and come back to your starting position. Perform all the trials at your comfortable speed, and you can look straight ahead or, if needed, look down to perform the trials". We explained the six experimental conditions to all the participants: no obstacle normal step width (NO-NORMAL), obstacle normal step width (OBS-NORMAL), no obstacle medial step width (NO-MEDIAL), no obstacle lateral step width (NO-LATERAL), obstacle medial step width (OBS-MEDIAL), and obstacle lateral step width (OBS-LATERAL). Participants performed three practice trials each of NO-NORMAL and OBS-NORMAL conditions first, where, NO-NORMAL condition was defined as the absence of any obstacle or mention of any stepping target line in the participants' walkway and OBS-NORMAL was defined as an obstacle placed from participant's starting toe position at 25% of their leg length and no mention of any stepping target line. We then calculated the step width of the first step from each trial of these two conditions by subtracting the frontal plane coordinates of the right heel marker after the first step from the left heel marker. We calculated the average step widths for the NO-NORMAL and OBS-NORMAL conditions. Medial and lateral target lines were placed on the laboratory floor for each participant with yarn and secured on either end with tape. To draw the medial target line for both NO-MEDIAL and OBS-MEDIAL, we drew a line running straight under the left foot for each participant. We drew lateral target lines for NO-LATERAL and OBS-LATERAL at doubled the distances of the average stepping widths of NO-NORMAL and OBS-NORMAL, respectively, the right side of the medial target line. All the participants were given three trials for each condition to practice. Participants performed five trials for each of these six conditions and completed 30 experimental trials in random order. We allowed our participants to see the walkway and obstacle if placed for the conditions throughout the whole time. A representative picture of a participant performing one of the experiment trials is shown in Figure 21.

Figure 21. Participant stepping on the 2nd force plate on the medial target line for the OBS-MEDIAL trial during the experiment. White line showing the medial target for NO-MEDIAL and OBS-MEDIAL. Yellow and red lines showing the lateral target lines placed on the floor using yarn for the NO-LATERAL and the OBS-LATERAL condition, respectively.

Data collection and analysis

We used a Vicon motion capture system using 11 infrared cameras operating at 100 Hz (Vicon Motion Systems, Oxford, UK) to calculate kinematic data and AMTI force plates (Advanced Mechanical Technology Inc., Newton, MA) to calculate kinetic data. To calculate kinematic data, we created a 12-segment rigid body model from 36 passive reflective markers placed over bony landmarks to identify segment length and track segment motion during stepping of the following segments: upper arms, lower arms, trunk, pelvis, upper legs, lower legs, and feet. The 2 AMTI force plates were placed successively to each other in the forward direction to collect kinetic data at 1000 Hz.

Kinematic and kinetic data were filtered with a 4th order low-pass Butterworth filter with a cutoff frequency of 6 Hz for kinematic and 12 Hz for kinetic data. To calculate whole-body CoM, CoM velocity, and trunk angle, we used 3-dimensional spatial coordinates of the markers placed on the body using Visual 3D software (C-Motion Inc., Rockville, MD, USA). We quantified the location of CoP, toe-off of the right foot, and right foot-strike we used ground reaction force data using Visual 3D software. We used custom MATLAB 2018a software to do further calculations. Step width and step time were calculated from the right heel and left heel data. Step time was calculated from the right foot toe-off to the right foot strike of the 1st step. For each trial, we calculated data from the 1st step taken. The distance between the CoM and the CoP position in the ML direction quantified ML CoM-CoP displacement, and we extracted peak value in the APA phase, value at the toe-off, and at maximum CoM-CoP displacement during swing for each trial. We extracted the peak value of ML trunk angle data as well as ML CoM velocity data at maximum CoM-CoP displacement during swing during the swing phase for each trial. We then averaged data across all trials in each condition to calculate the mean and standard deviation of all the variables of interest.

Statistical analysis

We used a 3-way mixed-effects repeated measures ANOVA (rmANOVA) with a group (fallers, non-fallers), as the between-subjects factor and step distance (no obstacle, obstacle) and step width (normal, medial, and lateral) as within-subject factors to compare means for peak ML CoM-CoP distance during APA, ML CoM-CoP distance at the toe-off, ML CoM-CoP distance at maximum CoM-CoP displacement during swing, peak ML trunk angle during swing, ML CoM velocity at maximum CoM-CoP displacement during swing. For descriptive purposes, we also ran 3-way rmANOVA on step length and step time (2 groups x 2 step distances x 3 step widths). All of these statistical analyses were conducted using SPSS software (version 25.0; IBM Corporation, Armonk, NY), and alpha was set at 0.05 for all the analyses. Bonferroni post-hoc adjustments were used to adjust for multiple comparisons.

Results

Step width, step length, and step time characteristics

All subjects showed modification of their first step width consistent with the intent. Step width characteristics are presented as mean and standard deviation (SD) in Table 6. Step width changed according to the step width tasks for both the groups (Table 6, $p < 0.001$). In the post-hoc analysis, step width was significantly greater in: 1) NO-NORMAL versus NO-MEDIAL ($p < 0.001$ for both groups), 2) NO-LATERAL versus NO-NORMAL ($p < 0.001$ for non-faller, $p = 0.005$ for faller), 3) OBS-NORMAL versus OBS-MEDIAL ($p < 0.001$ for both groups), and 4) OBS-LATERAL versus OBS-NORMAL (p

< 0.001 for non-faller, p = 0.024 for faller), showing differences in step widths across different step width tasks. Our methods did result in different step widths.

For step length, we found no significant main effect of group ($p > 0.05$); however, all participants took longer steps for the obstacle compared to the no obstacle condition ($p < 0.001$). For step length, we also observed a main effect of step width condition ($p = 0.002$) where participants reduced step length from normal to lateral ($p = 0.011$) and medial to lateral ($p < 0.001$) but not from normal to medial ($p > 0.05$) step widths. We found no significant 3-way interaction for group x step distance x step width ($p > 0.05$). There were no significant 2-way interactions for group x step distance, group x step width, and step distance x step width ($p > 0.05$ for all); however, there was a trend toward significance for group x step width interaction ($p = 0.050$). Table 7 shows mean and standard deviation of all the conditions for step length.

For step time, we observed significant main effects of step distance ($p < 0.001$) and step width ($p < 0.001$) conditions but not of group ($p > 0.05$). There was a significant interaction for group x step distance ($p = 0.022$) where for the obstacle, older non-fallers and fallers were different but not for the no obstacle condition. We also found a significant interaction for step distance x step width ($p = 0.03$). Post-hoc analysis showed that at the no obstacle condition step times increased/scaled from normal to medial ($p < 0.001$) and normal to lateral ($p = 0.001$) but not from medial to lateral ($p > 0.05$) step width conditions, however, at the obstacle condition step times scaled from normal to medial ($p = 0.009$), normal to lateral ($p < 0.001$), and medial to lateral ($p = 0.036$) step width conditions. No significant interaction was found for group x step distance x step width ($p > 0.05$). Table 7 shows mean and standard deviation of all the conditions for step time.

Table 6. Step width characteristics (mean ± SD) for both the groups

Variable	**Older non-fallers (n=12)**	**Older fallers (n=8)**
NO-NORMAL Step width (m)	0.14 ± 0.03	0.14 ± 0.02
NO-MEDIAL Step width (m)	0.00 ± 0.02	0.00 ± 0.01
NO-LATERAL Step width (m)	0.23 ± 0.07	0.24 ± 0.06
OBS-NORMAL Step width (m)	0.14 ± 0.04	0.15 ± 0.03
OBS-MEDIAL Step width (m)	0.00 ± 0.01	0.02 ± 0.02
OBS-LATERAL Step width (m)	0.25 ± 0.07	0.22 ± 0.08

Table 7. Step length and step time data for all the six conditions (mean ± SD)

	Non-fallers		**Fallers**	
Conditions	**Step length (m)**	**Step time (s)**	**Step length (m)**	**Step time (s)**
NO-NORMAL	0.54 ± 0.07	0.41 ± 0.04	0.53 ± 0.11	0.39 ± 0.04
NO-MEDIAL	0.57 ± 0.10	0.49 ± 0.06	0.51 ± 0.11	0.47 ± 0.05
NO-LATERAL	0.54 ± 0.10	0.47 ± 0.07	0.45 ± 0.12	0.46 ± 0.05
OBS-NORMAL	0.66 ± 0.06	0.71 ± 0.06	0.64 ± 0.07	0.79 ± 0.12
OBS-MEDIAL	0.65 ± 0.06	0.75 ± 0.09	0.63 ± 0.07	0.84 ± 0.15
OBS-LATERAL	0.64 ± 0.07	0.76 ± 0.09	0.60 ± 0.07	0.90 ± 0.15

Effect of task on ML CoM-CoP displacement during APA

Peak ML CoM-CoP displacement during APA showed no significant main effect of group ($p > 0.05$) but there were significant main effects of step distance ($p < 0.001$) and step width ($p = 0.004$) conditions. We observed a significant interaction for step distance x step width ($p = 0.02$) where for the no obstacle condition, ML CoM-CoP displacement was increased/scaled from normal to medial ($p = 0.005$) and decreased/scaled from normal to lateral ($p = 0.005$) and medial to lateral ($p < 0.001$) step width conditions but for the obstacle condition no scaling was observed for ML CoM-CoP displacement for any step width conditions ($p > 0.05$, Figure 22). We noted no significant interactions for group x step distance and group x step width ($p > 0.05$ for all). We also found no significant interaction for group x step distance x step width ($p > 0.05$).

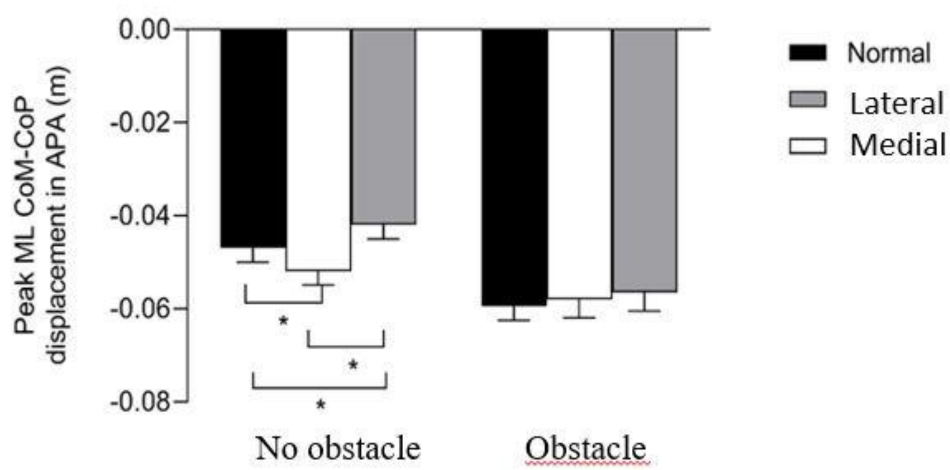

Figure 22. Peak ML CoM-CoP distance during APA for all the 4 conditions. Data is represented as mean ± SEM. * represents $p < 0.01$

Effect of task on ML CoM-CoP displacement at the toe-off

For ML CoM-CoP displacement at the toe-off, we observed no significant main effect of group ($p > 0.05$) but there were significant main effects of step distance ($p = 0.002$) and step width ($p < 0.001$) conditions. We found a significant interaction for step distance x step width ($p < 0.001$) where for the no obstacle condition ML CoM-CoP displacement at the toe-off was decreased/scaled from normal to medial ($p < 0.001$) and increased/scaled from medial to lateral ($p < 0.001$) step width conditions to a greater degree compared to the obstacle condition ($p = 0.003$ for both, Figure 23). We noted no significant interactions for group x step distance ($p > 0.05$) and only a trend toward significance for group x step width ($p = 0.078$). There was no significant interaction for group x step distance x step width ($p > 0.05$).

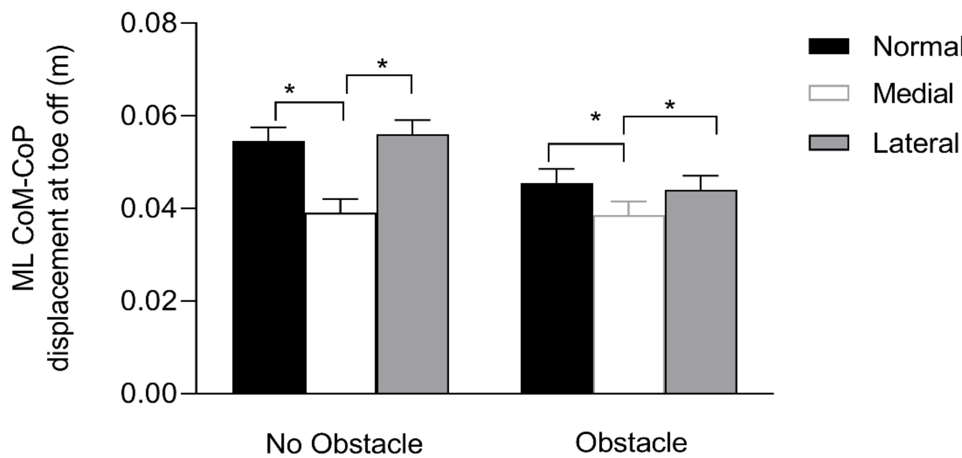

Figure 23. ML CoM-CoP distance at the toe-off for all the 4 conditions. Data is represented as mean ± SEM. *represents $p < 0.01$

Effect of task on ML CoM-CoP displacement during swing

ML CoM-CoP displacement at maximum CoM-CoP displacement during swing showed no significant main effects of group and step distance condition ($p > 0.05$ for all), but there was a significant main effect of step width condition ($p < 0.001$). We noted no significant interactions for group x step distance and step distance x step width ($p > 0.05$, for all). However, we found a significant interaction for group x step width ($p = 0.017$) where older non-fallers increased/scaled ML CoM-CoP displacement at maximum CoM-CoP displacement during swing to a greater degree from normal to lateral ($p < 0.001$) and medial to lateral ($p < 0.001$) step widths than fallers ($p = 0.040$ and $p < 0.001$), whereas the degree of scaling was similar from normal to medial ($p < 0.001$) for both the groups (Figure 24). We observed no significant interaction for group x step distance x step width ($p > 0.05$). An example of individual data points of ML CoM-CoP displacement during swing for the OBS-LATERAL condition is shown in Figure 25.

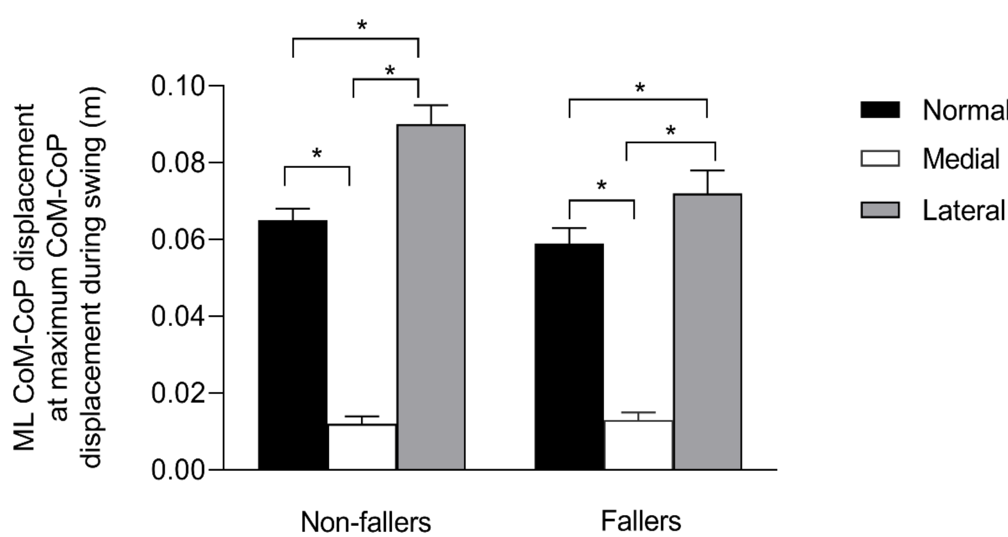

Figure 24. ML CoM-CoP distance at maximum CoM-CoP displacement during swing for significant group x step width interaction. Data is represented as mean ± SEM. *represents $p<0.05$

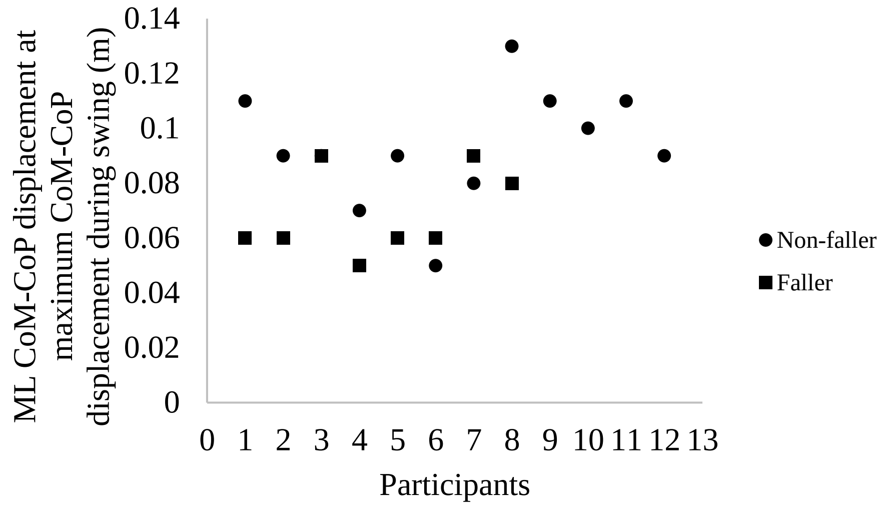

Figure 25. Individual data points of ML CoM-CoP displacement during swing performing the OBS-LATERAL condition. Circles represent data of non-fallers group and squares represent data of the faller group.

Effect of task on trunk angle and CoM velocity

Peak ML trunk angle during swing showed no significant main effects of group, step distance, and step width ($p > 0.05$ for all). We observed no significant interactions for group x step width and step distance x step width ($p > 0.05$, for all). However, we found a significant interaction for group x step distance ($p = 0.033$). Post-hoc analysis showed that older non-fallers increased peak ML trunk angle during swing from the no obstacle to the obstacle ($p = 0.008$) condition, but older fallers ($p > 0.05$, Figure 26) maintained peak ML trunk angle between the two step distance conditions. There was no significant interaction for group x step distance x step width ($p > 0.05$). An example of individual data points of peak ML trunk angle for the OBS-LATERAL condition is shown in Figure 27.

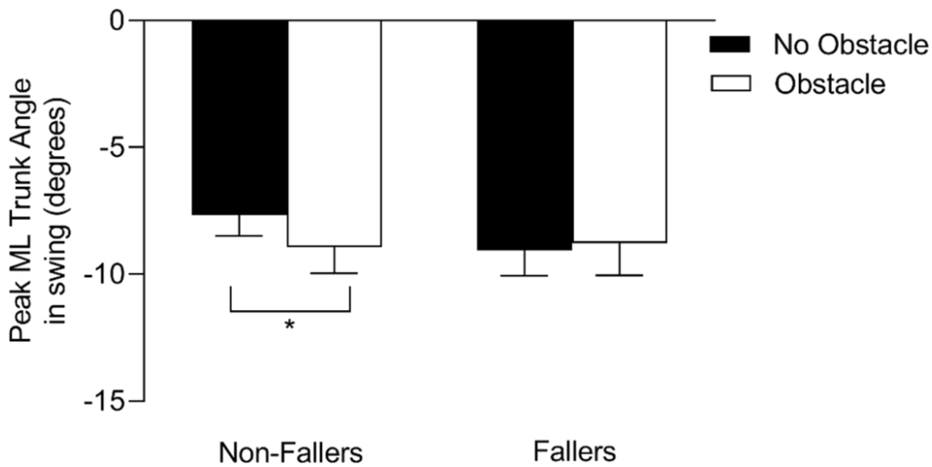

Figure 26. Peak ML trunk angle during swing for significant group x step distance interaction. Negative numbers indicate the trunk tilts toward the stance side. Data is represented as mean ± SEM. *represents $p < 0.05$

For ML CoM velocity at maximum CoM-CoP displacement during swing, we found no significant main effect of group ($p > 0.05$), but there were significant main effects of step distance and step width ($p < 0.001$ for all). We also found no significant interactions for group x step distance and group x step width ($p > 0.05$, for all). However, we found a significant interaction for step distance x step width ($p < 0.001$). Post-hoc analysis showed that ML CoM velocity at maximum CoM-CoP displacement during swing was increased/scaled to a similar degree from normal to lateral ($p < 0.001$) for both the no

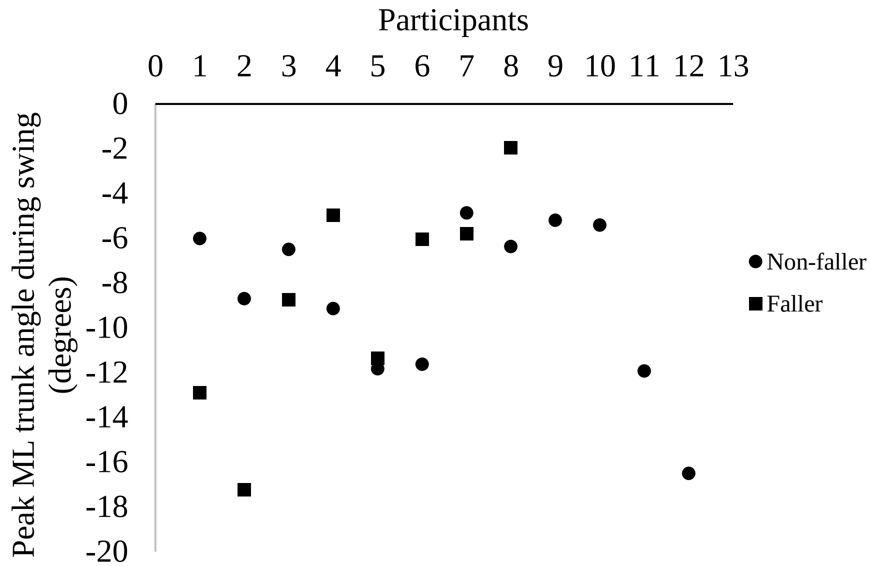

Figure 27. Individual data points of peak ML trunk angle during swing performing the OBS-LATERAL condition. Circles represent data of non-fallers group and squares represent data of the faller group.

obstacle and the obstacle conditions. In contrast, in the obstacle condition decreased/scaled from normal to medial ($p < 0.001$) and increased/scaled from medial to lateral ($p < 0.001$) step widths to a greater degree compared to the no obstacle condition ($p < 0.001$, Figure 28). We observed no significant interaction for group x step distance x step width, but a trend toward significance was noted ($p = 0.070$).

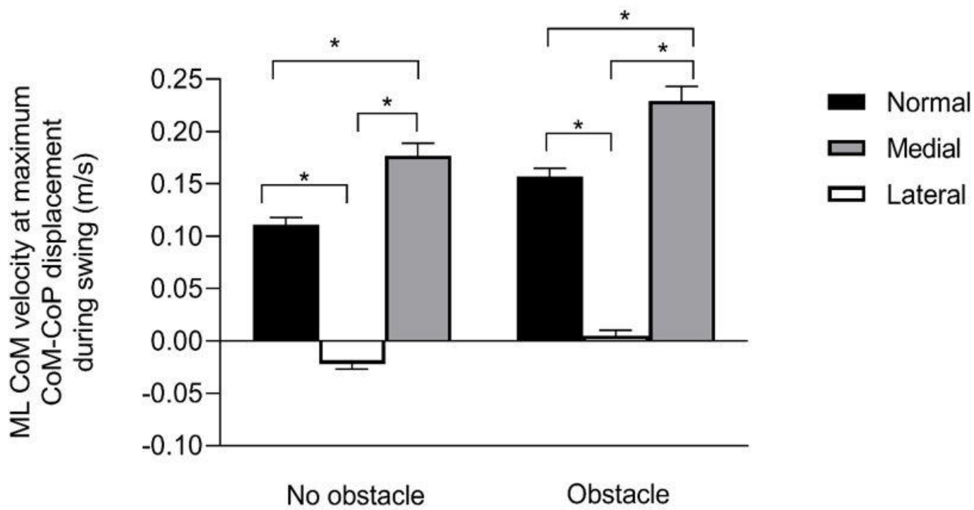

Figure 28. ML CoM velocity at maximum CoM-CoP displacement during swing for all the 4 conditions. Data is represented as mean ± SEM. *represents where $p < 0.001$

Discussion

The purpose of this study was to compare the scaling of postural control variables in the ML plane between older non-fallers and fallers when stepping with increased and decreased step widths during GI. The major findings of our study do not support our initial hypothesis. Instead, we found that:

1) at maximum CoM-CoP displacement during swing, irrespective of step distance conditions, older fallers decreased scaling of ML CoM-CoP displacement compared to non-fallers;

2) fallers do not lean toward the stance limb differently between the no obstacle and the obstacle condition whereas non-fallers do;

3) irrespective of fall history, older adults for the no obstacle condition: a) increased peak ML CoM-CoP displacement during APA in medial step width and b) reduced ML CoM-CoP displacement in lateral step width, compared to the normal step width. No differences between the step width tasks for the obstacle condition were noted;

4) older adults irrespective of their fall history reduced ML CoM-CoP displacement at the toe-off for the medial compared to the normal step width for the no obstacle to a greater degree than the obstacle condition; and

5) older adults irrespective of their fall history increased ML CoM velocity at maximum CoM-CoP displacement during swing in lateral compared to normal step width task and decreased in medial compared to normal step width task to a greater degree for the obstacle than the no obstacle condition.

Depending on the phase of GI before the toe-off, older adults (both non-fallers and fallers) employed different approaches for medial versus lateral step width tasks (irrespective of step distance conditions) to modulate ML CoM-CoP displacement. Whereas, during the swing phase, older fallers compared to non-fallers, scale ML CoM-CoP displacement conservatively for both medial and lateral step width tasks irrespective of the step distance conditions. Also, our results suggest that older adults reduce the scaling of ML CoM-CoP displacement when an obstacle is present. This may indicate a cautious behavior undertaken by older adults to potentially reduce lateral instability while crossing the obstacle. With lateral trunk tilt data, our findings also reflect that fallers adopt a compensatory approach while modulating peak ML trunk angle for the obstacle clearance compared to non-fallers. Another finding of our study suggests that older adults adopt their ML CoM velocity during the swing phase in the task- and direction-specific manner.

Stability during APA

CoM-CoP displacement is directly proportional to CoM acceleration [66]. In older adults, increased peak ML CoM-CoP displacement during APA in medial step width could serve to provide adequate acceleration to the leg to finish the GI task successfully, positioning their swing limb more medially than during normal gait. The presence of larger APA in the ML direction suggests an increased role of predictive control to achieve ML stability in young adults [106]. Similarly, in our study, increased peak ML CoM-CoP displacement during APA in the NO-MEDIAL condition could indicate an important role of predictive control to achieve stability during the end of the GI. Decreased peak ML CoM-CoP displacement during APA in the lateral step width task could serve to avoid ML CoM overshoot (instability) due to increased momentum while reaching for the lateral target. However, in the obstacle condition, for successful task completion, older adults do not scale peak ML CoM-CoP displacement during APA differently for different step width tasks as there is an increased risk of hitting the obstacle when performing either medial or lateral step width task.

Our study suggests peak ML CoM-CoP displacement during APA might not be adequately discriminant to differentiate between fallers and non-fallers. Our findings are similar to results from a previous study found during choice stepping reaction task ML APA shift did not differ between older fallers and non-fallers [14]. Our study findings of no difference in peak ML CoM-CoP displacement between fallers and non-fallers is consistent with this prior study's findings.

Our results suggest that older adults in anticipation of potential ML instability imposed by obstacle constraint reduce chances of instability by modulating ML APA. In previous studies, it has been suggested that larger ML CoP during APA in the obstacle condition during GI is to ensure stability at heel strike in young adults [103,106]. In young adults, with obstacle crossing, swing duration increases and may increase lateral instability necessitating larger ML APA to maintain stability [103,106]. Similarly, in our study, we found with older adults that for the obstacle condition, peak ML CoM-CoP displacement during APA was increased, which could be related to a prolonged swing duration for the obstacle crossing.

Stability at the toe-off

Irrespective of obstacle conditions, older adults scale ML CoM-CoP displacement at the toe-off from normal to medial step width probably due to task requirement as the CoM had to move toward the stance limb for stepping on the medial target. However, ML CoM-CoP displacement at toe-off is not changed in lateral compared to normal step width, irrespective of obstacle conditions, possibly to avoid ML CoM overshoot. There is evidence that with increased step width during GI in young adults, there is an incidence of ML overshoot of the CoM position [90]. Similarly, with older adults in our study, a wide lateral step width (i.e., in the lateral step width condition) may increase ML CoM overshoot, which may further increase challenges that reported for some older adults to maintain ML stability [111].

Trips are common in older adults while stepping over obstacles [8]. Stepping over obstacles with different step widths can induce instability, and hence trips in older adults. In young adults, it has been suggested that with increased duration in the swing phase during obstacle crossing, a larger CoM shift toward the swing limb side can occur [106]. In our study, with increased step time in the obstacle compared to the no obstacle conditions, older adults could have decreased ML CoM-CoP at toe-off potentially due to reduced ML stability associated with aging [111]. However, this also reflects a counter strategy, when crossing an obstacle, to prevent increased CoM shift toward the lateral side at the heel strike that occurs at the end of GI. Scaling of ML CoM-CoP displacement at the toe-off from the normal to medial step width was observed to a lesser degree in the obstacle than the no obstacle conditions emphasizing that obstacle presence may pose a challenge to the stepping task above and beyond the targeted lateral steps. Overall, it seems older adults adopt a conservative approach to decrease the chances of potential instability by scaling ML CoM-CoP displacement less in the obstacle compared to the no obstacle condition.

It has also been suggested previously that fallers increase stability at toe-off compared to non-fallers for a lateral target but not for a central target during choice reaction stepping task [14]. However, in our study, we did not find a statistical difference between fallers and non-fallers for our stability marker, potentially due to a few differences between our study and the Tisserand et al. (2016) study. First, in our task, we did not ask our participants to step “as quickly as possible”. Second, we used CoM-CoP displacement as the marker of stability, whereas the margin of stability was used in their study. Finally, we compared normal to lateral step width rather than comparing central to lateral step width.

Stability during swing

Based on our study findings, it seems fallers can be differentiated from non-fallers using ML CoM-CoP displacement at maximum CoM-CoP displacement during swing in a scaled ML step width task during GI. In a previous study, reduced ML CoP displacement was suggested to differentiate fallers from non-fallers in dynamic postural control task [112]. Compared to non-fallers, fallers have weak hip abductors and adductors [94] and diminished lateral stability [111]. Such differences may partially explain how fallers in our study reduced the scaling of ML CoM-CoP displacement at maximum CoM-CoP displacement during swing compared to non-fallers. In the lateral step width task, fallers reduced the scaling of ML CoM-CoP displacement at maximum CoM-CoP displacement during swing compared to non-fallers. Increased stride width is associated with fear of falling in gait [113]. It is possible that in our study because of increased fear of falling and to avoid potential instabilities associated with task completion, ML CoM-CoP displacement at maximum CoM-CoP displacement during swing was reduced in fallers in the lateral step width condition compared to non-fallers. Moreover, increased ML CoM-CoP displacement at maximum CoM-CoP displacement during swing in the medial step width condition and decreased ML CoM-CoP displacement in the lateral step width condition may require increased postural control demand, and hence increased muscle strength in older adults. Therefore, older adults irrespective of fall status, adjusted ML

CoM-CoP displacement at maximum CoM-CoP displacement during swing in the medial step width condition based upon task requirements.

In a previous study, it was reported that in older adults, the CoM velocity decreased with reduced step width during step initiation [90], as participants may hesitate to increase velocity as it may increase fall risk to the side. Data from our study are consistent with these results. However, in older adults, increased ML CoM velocity in the lateral step width task compared to normal step width task may have occurred possibly to aid successful task completion, that is, to not undershoot the target.

Tilting the trunk laterally in older adults might put them at risk of instability. Interestingly, we found that older non-fallers increase the trunk tilt toward the stance limb while crossing obstacle versus no obstacle, which may be necessary to perform obstacle crossing successfully. Although increased trunk motions can contribute to further destabilization of postural stability with aging [114,115], counterintuitively non-fallers increased their trunk tilt for the obstacle condition toward the stance limb. Increased trunk tilt toward the stance limb may help to successfully perform the task at the cost of creating potential instability or to bring the CoM closer to the stance limb. A previous study shows that young adults tilt their trunk laterally toward the swing limb during the swing phase of GI [34]. In our study, older adults tilt the trunk laterally toward the stance limb during GI. This may be a compensatory strategy to account for weak hip muscles in older adults. Further, a lack of scaling response for the trunk tilt for the obstacle versus the no obstacle condition in older fallers may reflect a conservative approach or a cautious approach due to: a) increased fear of falling, b) hip muscle weakness, c) impaired depth perception on an obstacle and/or, d) impaired proprioception [99,116,117]. Greater control of trunk versus foot in fallers during gait has been reported before [118]. Though we did not find differences between fallers and non-fallers for different step width tasks, data from our study indicates that, the trunk was constrained in older adults. This may have been a result of a compensatory approach, which was shown by the lack of a difference in sideways trunk lean when foot placement was changed (i.e., medial versus lateral step width tasks).

It is well established that older adults reduce step length compared to young adults during GI [15,16]. A reduced step length in older versus young adults could be suggestive of the requirement of proper foot placement and maintain stability at foot strike. In our study with older adults, we found larger step length reduction in a lateral but not in a medial step width task when compared to a normal step width task. In a previous study with older adults, swing duration was reduced when performed stepping on a lateral target compared to a central target, whereas in our study, we found step time increased from normal to lateral. It could be because of differences in the nature of the task in terms of temporal constraint or because of the central target, which was not a normal stepping target [14]. For the obstacle condition, older non-fallers and fallers were different (step time increased in fallers) but not for the no obstacle condition. Inherent postural instability in fallers may encourage a larger safety margin to avoid hitting an obstacle and increase their swing time to clear the obstacle.

Limitations

A technical limitation of this study was the limited width of the floor-embedded force plates. Specifically, medial and lateral step width target lines were placed on the floor based on normal stepping during practice trials. Four participants, all fallers, had lateral step widths that fell beyond the width of our force plates. The lack of hip strength data was another limitation of this study. Availability of the hip muscle strength data such as hip abductors strength could have helped understand the trunk tilt toward the stance limb, perhaps, in a better-informed way. We always asked the participants to start the experimental tasks with their right leg. Always starting with the same right leg may introduce bias or variability for those subjects that may have preferred to start with their left leg. However, it is in line with a previous study that consistently used the right leg to test the obstacle crossing task during GI [51]. Further, we undertook 40 trials for our experimental tasks; however, this large number of trials in older adults may have turned the task boring and therefore, might have affected their performance. Also, a standard height of obstacle was used in our study, which may have affected how fallers and non-fallers scale the dependent variables of our study and could limit the generalizability of our findings. Stance width was also fixed based on participants' practice trials stance width, and hence different stance width of participants has the potential to affect our results and generalizability of our findings.

Conclusion

In older adults, impaired ML stability is associated with an increased risk of falling [119,120]. Our results suggest that for our ML step width task, ML CoM-CoP displacement in the swing phase differentiates fallers and non-fallers. It is interesting that before the toe-off, depending on the phase of GI older adults irrespective of fall status, employ the approach differently for medial versus lateral step width tasks to modulate ML CoM-CoP displacement. However, during the swing phase, older fallers compared to non-fallers, scale ML CoM-CoP displacement conservatively for both medial and lateral step width tasks irrespective of the obstacle conditions. Moreover, during the swing phase, ML CoM velocity changes according to task constraint and task direction, but the trunk lateral tilt mainly changes according to obstacle constraint. Overall, our findings imply that the effects of aging on sensorimotor processing and musculoskeletal decline make scaled step width GI task more difficult for the postural control system to optimally solve the complex motor task underlying standing and walking.

CHAPTER 5

CONCLUSION

Review of Specific Aims

The overall goal of this dissertation was to investigate the effect of age and fall status on the scaling of postural control variables when challenged with scaled tasks during gait initiation (GI). For scaling task difficulty, we utilized two ways to study the behavior of postural control variables– scaling step length (anterior/posterior) and scaling step width (medial/lateral). Older adults report increased falls during activities that require a transfer of body weight during changes in their base of support. Our results show that GI and obstacle crossing are two such activities that require a proper transfer of weight to avoid postural instabilities and fall in older adults. The following specific aims guided our work:

Specific Aim 1. To characterize postural control in a scaled GI task in the presence of spatio-temporal constraints imposed by an obstacle in healthy adults.

Specific Aim 2. To compare the scaling of the postural control variables between older non-fallers and fallers to a scaled step distance task during GI in the presence of spatio-temporal constraints imposed by an obstacle.

Specific Aim 3. To determine the extent to which older fallers and non-fallers scale the postural control variables when stepping over an obstacle while increasing or decreasing step width to challenge medial-lateral (ML) stability.

The hypothesis for Aim 1 was that in response to increased step distance when crossing obstacles, healthy adults will scale the CoM-CoP displacement with increasing postural disturbance during GI. The study findings fully support this hypothesis. We looked at the scaling of CoM-CoP displacement and trunk angle at maximum CoM-CoP displacement during swing in the anterior-posterior (AP) direction in response to stepping over an obstacle placed at progressive distances during GI in healthy adults. We found that healthy adults scale both the CoM-CoP displacement and trunk angle at maximum CoM-CoP displacement during swing in the AP direction. It was interesting to observe that CoM-CoP displacement at maximum CoM-CoP displacement during swing in the AP direction increases with increasing obstacle distances. However, we also observed that the trunk angle at maximum CoM-CoP displacement during swing in the AP direction decreases with increasing obstacle distances. Looking at the relationship between these variables, we found a positive association for all the conditions, whether no obstacle, near obstacle, or far obstacle distance. These results reflect that healthy adults actively modify CoM-CoP displacement per task constraints and trunk angle to reduce gravitational forces.

The hypothesis for Aim 2 was that during GI, compared to older non-fallers, fallers will not scale their posture control, and that the fallers will not show scaling of 1) peak AP CoM-CoP displacement during APA and 2) AP CoM-CoP displacement at the toe-off and, 3) AP CoM-CoP displacement at maximum CoM-CoP displacement during swing while performing progressive stepping distances resulting in group x step distance interaction. Our hypothesis was partially supported by the study findings. We compared the scaling of the postural control variables between older non-fallers and fallers in response to scaled stepping distance tasks with obstacle as constraint during GI. We found a significant interaction between group and condition for CoM-CoP displacement at the toe-off only.

Older fallers reduce peak AP CoM-CoP displacement during APA irrespective of conditions. However, older fallers also reduce AP CoM-CoP displacement at the toe-off compared to non-fallers but only at the middle and far obstacle distance. Our secondary analysis revealed that older fallers increase trunk forward lean compared to non-fallers. These findings reflect the conservative approach that older fallers adopt prior to the toe-off, probably to counter postural instability associated with task performance during the swing phase of the GI. Older fallers may have increased control of the CoM movement during the preparatory phase in anticipation of increased instabilities in the swing phase, which could have arisen by the increased trunk forward lean while successfully completing the challenging tasks.

The hypothesis for Aim 3 was that older fallers performing a GI task with targeted initial step widths would result in no scaling of 1) peak ML CoM-CoP displacement during APA and 2) ML CoM-CoP displacement at the toe-off and, 3) ML CoM-CoP displacement at maximum CoM-CoP displacement during swing compared to non-fallers resulting in group x step distance x step width interaction. The study findings do not fully support our hypothesis. We compared postural control variables between older non-fallers and fallers in response to scaled step width tasks during GI. We found no significant group x step distance x step width interaction. No fall status-related differences existed for the scaling of ML CoM-CoP displacement during APA or at the toe-off for different step width tasks. However, older fallers scaled less than non-fallers during the swing phase, implying a strategy to reduce active postural control demand in fallers. It can also be inferred that medial step width task was more challenging to perform successfully during the swing phase of GI for the fallers. Additional findings of no scaling of the lateral trunk tilt toward the stance limb irrespective of obstacle presence highlight the approach that older fallers adopt to complete the task successfully compared to non-fallers.

Strengths of the studies

The study findings of Aim 1 add to the body of literature in understanding postural control responses to the increasing level of challenges when imposed by an obstacle constraint during GI in young adults. This project design can be used as a framework to investigate postural imbalance in other clinical populations with gait issues.

The study findings of Aim 2 are important as they provide insight into postural stability responses in older fallers during challenging tasks in the AP direction during GI. Moreover, the results of this study may lead to the development of novel rehabilitation based targeted intervention for balance impaired individuals in the future.

The study findings of Aim 3 contribute to gain knowledge about the scaling of postural control variables in the ML, the direction in which falls mostly occur in older adults. The results from Aim 3 also help better understand the behavior of older fallers to complete the task successfully when challenged with varied difficulties. This study contributes to gaining new knowledge about the scaling of postural control variables in the ML direction in which falls occur majorly in older adults and hence help better understand the behavior of fallers to complete the task successfully.

Limitations

There are several limitations of this dissertation work. One limitation of this dissertation work was that we asked our participants to use their right leg to initiate stepping in all the experiments. It could introduce bias or variability for those subjects that may have preferred to start with their left leg. Another limitation of this work is the lack of measurement of hip muscle strength, a possible confounder for changes in trunk angle that were noted. Future work would benefit from the inclusion of muscle strength testing. Although we used a standard method [14,94,121] to define and recruit fallers in our study, we did not find expected differences between faller and non-faller groups on certain variables (such as AP CoM-CoP displacement at maximum CoM-CoP displacement during swing, peak ML CoM-CoP displacement during APA). Berg Balance Scale, Timed-Up and Go test, and Activities-Specific Balance Confidence Scale scores were not significantly different between fallers and non-fallers in our studies. Another criterion for faller and non-faller groups may have resulted in more group differences, though it is possible that fallers and non-fallers may not have differed similar to our study findings.

Another limitation of my dissertation was the use of the standard height of the obstacle for all the participants, as it may have affected the generalizability of our findings. The standard obstacle height may have been relatively too high/low for individuals with shorter/longer than average leg lengths, presenting an easier/harder task. The average subject height was 166 cm, and none of the participants were outliers, indicating this most likely had a minor to no impact.

Probably, the ML scaling task was not scaled enough to elicit postural response differently between fallers and non-fallers. The study had a technical limitation, the limited width of the floor-embedded force plates, which led to a loss of 4 participants' data in our study. A larger force plate, particularly in the ML direction, could provide a greater challenge as the medial and lateral targets would be further apart, and possibly more significant changes in postural control variables between fallers and non-fallers.

We looked at the postural control variables during the 1st step of GI only. Investigating two steps instead of just 1st step might have helped to better inform postural control variables behavior in a certain manner in our studies such as changes in the trunk angle during 1st step could be related to step width in the consecutive step. For our studies, we mainly examined variables such as CoM-CoP displacement, trunk angle, and CoM velocity. Although we did not analyze variables such as variability of CoM-CoP displacement and trunk angle and variables in non-task specific directions, these can be examined in detail as the secondary analysis of my dissertation.

Key findings

The overall objective of this work was to understand how older adults classified as fallers and non-fallers alter their balance control strategies during postural challenging GI tasks. To the best of our knowledge, our work is the first report to determine age and fall status related effects on the scaling of AP and ML postural control variables across scaled GI tasks. We used a novel paradigm, a scaled step distance task and step width task that created conditions for AP and ML instability during GI while stepping over an obstacle. The central finding of my dissertation is that the successful completion of scaled tasks requires scaling of the postural control variables and that the scaling is different between fallers and non-fallers as well as during different events of GI. In general, older fallers compared to non-fallers seemingly adopt a more conservative approach, reduced CoM-CoP displacement either before or after the toe-off depending on the scaled tasks, for successful completion of all the tasks. It is possible that older fallers might be adopting a conservative approach as a strategy to gain a biome chanical advantage by keeping CoM closer to the BOS. Reduced CoM-CoP displacement in fallers could help control horizontal CoM acceleration to avoid instability at the toe-off or the heel contact during GI. In the AP scaled GI task, faller and non-faller groups scaled differently in the AP direction **prior** to the toe-off. However, in the ML scaled GI task, ML scaling differences between the faller and non-faller groups occurred **after** the toe-off of the GI. It could be assumed that the in AP scaled tasks CoM movement is more tightly controlled and maintained closer to the BOS by reducing AP CoM-CoP displacement prior to the toe-off, but in ML scaled tasks tighter control of CoM movement by reduced ML CoM-CoP displacement is seen after the toe-off. Such a strategy of reduced CoM-CoP displacement in fallers could also have an unintended consequence, such as an increased energy cost in trying to keep CoM and CoP closer to each other. Decreasing CoM-CoP displacement could require an increased muscular activity; thus, necessitating an increase in metabolic demand and an increase in energy cost to keep CoM and CoP closer to each other. On the other hand, it is also probable that due to inherent poor postural stability in fallers as evident in the literature, fallers try to keep themselves stable by reducing CoM-CoP per the directionality of the task. Functionally, increased fall-risk in fallers could be associated with attempts to reduce active postural control demands associated with increased CoM-CoP displacement. Hence, in fallers, we note reduced scaling of CoM-CoP displacement compared to non-fallers.

It was also interesting to find that healthy adults represent healthy postural control regulation by scaling postural control variables per the task demand. This finding can be used by scientists/clinicians/rehabilitation professionals to examine the scaling of postural control variables in balance-impaired individuals compared to a standard control. Moreover, it seems that CoM-CoP displacement and trunk lean relationships are important for understanding active and passive postural control regulation. In older fallers, a negative association between peak AP CoM-CoP displacement during APA and peak trunk lean during swing with AP scaled task perhaps is a reflection of coupling across perceptual (using more forward trunk lean to get a better view of the obstacle) as well as musculoskeletal systems (reducing CoM-CoP displacement by using differential muscle activation). We also noted that in young adults, there is coordination between the upper and lower body segments, as seen with a positive association between CoM-CoP displacement and trunk lean with AP scaled GI task. Looking at a broader comparison between healthy adults (less than 40 yrs of age) and older adults (above 65 yrs of age) with the scaled step distance tasks during GI, it seems that older adults prefer a cautious strategy based on CoM-CoP displacement at maximum CoM-CoP displacement during swing than healthy adults. This could perhaps be a reflection of inflexibility in older adults, especially in fallers with advancing age compared to healthy adults. Inflexibility in older fallers could arise because of multitude of reasons such as tight or stiff muscles or capsular tightness that could restrict joint movement. Interestingly, among older adults, in fallers, compared to non-fallers looking at the increased trunk forward lean during scaled AP GI tasks could indicate the flexibility of the neuromotor system of fallers. Notably, increased trunk forward lean in fallers could implicate a decrease of stability and could be one of the factors contributing to their previous fall(s). Largely, older adults (whether faller or non-faller) show flexible behavior in relation to trunk motions in the AP direction (increased trunk forward lean) in response to scaled AP GI tasks; for the ML scaled GI task, older adults show inflexible behavior in relation to trunk motion in the ML direction (unchanged trunk tilt) in response to scaled ML GI tasks.

The operational definition of scaling of my dissertation is not dependent on the number of levels of differences. Rather, scaling was dictated by a statistically significant difference obtained by within-subjects effects via SPSS statistical analysis. Thus, the dissertation findings showed that scaling occurred irrespective of the number of levels of significant differences found. Based on the specific Aims 2 and 3, my definition of scaling did not allow me to look at within-group differences if no significant interaction was found. Thus, it limited the broader understanding of the effect of increased task difficulty at each level on postural control variables within the groups. However, in Aim 1, my operational definition of scaling did not affect outcomes because there was only one group.

Findings from my dissertation can be used as a diagnostic and prognostic tool to assess postural instability and effectiveness of balance rehabilitation, respectively, in older fallers. Gait initiation is one of the most common activities of daily living in ambulatory adults. Notably, my dissertation findings relay the importance of testing GI with an obstacle as a constraint to identify postural instability in clinical populations with balance issues. This is important because very few balance tests exist that focus on GI.

Future research

The findings of this work could be of great interest when designing interventions in older adults with a history of falls in the past 12 months. However, prior to the generalization of our results, it will be helpful to study older fallers who would have BBS score significantly lower than non-fallers to give us more insight into balance issues of fallers performing these tasks. Further examination should look at variability in fallers and non-fallers while performing scaled GI tasks to get a deeper understanding of their postural control regulation. For future steps, recruiting more participants would increase statistical power, where we found a trend toward significance. Future work should consider collecting data when initiating with both limbs to determine if the leg preference affected the results. Also, including an equal sample size of older fallers and non-fallers, would be helpful to draw a more accurate conclusion of this work. To further understand the deficits of older fallers in maintaining postural stability, more investigations will be needed that challenge postural stability under specific and varied conditions. Also, changing temporal constraints using reaction time tasks will be interesting to observe the behavior of older fallers and non-fallers. Defining scaling for significant differences between each level of task difficulty could inform about scaling behavior in response to tasks of increasing difficulty at a greater depth. Studying flexibility in older adults, especially in fallers with scaled GI tasks, could reveal underlying factors affecting postural stability. Finally, given the results of this dissertation, it will be important to recruit and test older adults with a clinical balance impairment to identify the stability risk factors that increase the likelihood of instability.

This work is the first attempt to examine the effects of scaled step distances and step widths using obstacle as a constraint during GI in older non-fallers and older fallers. From this study, it is worth noting that older adults, whether fallers or non-fallers, performed the tasks successfully without hitting the obstacle even for the far obstacle distance. A novel experimental design was implemented in this project that can be used to better understand the problems faced during activities of daily living in older adults with a balance issue. To conclude, older fallers adopt a conservative approach to successfully perform the task and show reduced scaling compared to non-fallers. This work suggests that the scaled GI tasks should be accounted to assess stability in clinical research. Scaling in different phases of GI can give us a tool to investigate balance impairment. This work helps to understand the postural control behavior of young, older non-fallers and fallers adults using scaled GI tasks. Overall, this study can inform improvement in mobility and quality of life in older individuals.

REFERENCES CITED

[1] S.W. Muir, K. Berg, B. Chesworth, N. Klar, M. Speechley, Quantifying the magnitude of risk for balance impairment on falls in community-dwelling older adults: a systematic review and meta-analysis, J. Clin. Epidemiol. 63 (2010) 389–406. doi:10.1016/j.jclinepi.2009.06.010.

[2] B.E. Bergen G, Stevens MR, Falls and fall injuries among adults aged [greater than or equal to] 65 years, 2014.

[3] E.R. Burns, J.A. Stevens, R. Lee, The direct costs of fatal and non-fatal falls among older adults — United States, J. Safety Res. 58 (2016) 99–103. doi:10.1016/j.jsr.2016.05.001.

[4] T.R. Englander F, Hodson TJ, Economic dimensions of slip and fall injuries, J. Forensic Sci. 41 (1996) 733–46.

[5] I. Melzer, N. Benjuya, J. Kaplanski, Postural stability in the elderly: A comparison between fallers and non-fallers, Age Ageing. 33 (2004) 602–607. doi:10.1093/ageing/afh218.

[6] C. Toulotte, A. Thevenon, E. Watelain, C. Fabre, Identification of healthy elderly fallers and non-fallers by gait analysis under dual-task conditions, Clin. Rehabil. 20 (2006) 269–276. doi:10.1191/0269215506cr929oa.

[7] B. Müjdeci, S. Aksoy, A. Atas, Evaluation of balance in fallers and non-fallers elderly, Braz. J. Otorhinolaryngol. 78 (2012) 104–109. doi:10.5935/1808-8694.20120016.

[8] S.N. Robinovitch, F. Feldman, Y. Yang, R. Schonnop, P.M. Luen, T. Sarraf, J. Sims-gould, Video Capture of the Circumstances of Falls in Elderly People, Lancet. 381 (2013) 778–782. doi:10.1016/S0140-6736(12)61263-X.Video.

[9] G.S. Tinetti ME, Speechley M, Risk Factors for Falls among Elderly Persons Living in the Community, N. Engl. J. Med. 319 (1988) 1701–1707.

[10] S. JH, The effect of age on the control of sway, Gerontol. Clin. (Basel). 5 (1963) 129–38.

[11] M.D.G. Christopher P. Hurt, Age-related differences in the maintenance of frontal plane dynamic stability while stepping to targets, Physiol. Behav. 176 (2017) 139–148. doi:10.1016/j.physbeh.2017.03.040.

[12] A.F. Polcyn, L.A. Lipsitz, D.C. Kerrigan, J.J. Collins, Age-related changes in the initiation of gait: Degradation of central mechanisms for momentum generation, Arch. Phys. Med. Rehabil. (1998). doi:10.1016/S0003-9993(98)90425-7.

[13] Y. Jian, D. Winter, M. Ishac, L. Gilchrist, Trajectory of the body COG and COP during initiation and termination of gait, Gait Posture. 1 (1993) 9–22. doi:10.1016/0966-6362(93)90038-3.

[14] R. Tisserand, T. Robert, P. Chabaud, M. Bonnefoy, L. Chèze, Elderly fallers enhance dynamic stability through anticipatory postural adjustments during a choice stepping reaction time, Front. Hum. Neurosci. 10 (2016) 1–15. doi:10.3389/fnhum.2016.00613.

[15] G.A. Mbourou, Y. Lajoie, N. Teasdale, Step length variability at gait initiation in elderly fallers and non-fallers, and young adults, Gerontology. 49 (2003) 21–26. doi:10.1159/000066506.

[16] B.C. Muir, S. Rietdyk, J.M. Haddad, Gait initiation: The first four steps in adults aged 20-25 years, 65-79 years, and 80-91 years, Gait Posture. 39 (2014) 490–494. doi:10.1016/j.gaitpost.2013.08.037.

[17] S. Park, F.B. Horak, A.D. Kuo, Postural feedback responses scale with biomechanical constraints in human standing, Exp. Brain Res. 154 (2004) 417–427. doi:10.1007/s00221-003-1674-3.

[18] A.L. Adkin, J.S. Frank, M.G. Carpenter, G.W. Peysar, Postural control is scaled to level of postural threat, Gait Posture. 12 (2000) 87–93. doi:10.1016/S0966-6362(00)00057-6.

[19] E. Yiou, R. Artico, C. Teyssedre, O. Labaune, P. Fourcade, Anticipatory Postural Control of Stability During Gait Initiation Over Obstacles of Different Height and Distance Under Reaction-Time and Self-Initiated Instructions, in:

Neuroergonomics, Elsevier, 2018: pp. 233–234. doi:10.1016/b978-0-12-811926-6.00051-8.

[20] A.E. Patla, A. Adkin, T. Ballard, Online steering: Coordination and control of body center of mass, head and body reorientation, Exp. Brain Res. 129 (1999) 629–634.

[21] P.M. McAndrew Young, J.B. Dingwell, Voluntarily changing step length or step width affects dynamic stability of human walking, Gait Posture. 35 (2012) 472–477. doi:10.1016/j.gaitpost.2011.11.010.

[22] L.S. Chou, K.R. Kaufman, R.H. Brey, L.F. Draganich, Motion of the whole body's center of mass when stepping over obstacles of different heights, Gait Posture. 13 (2001) 17–26. doi:10.1016/S0966-6362(00)00087-4.

[23] E.L. Stegemller, T.A. Buckley, C. Pitsikoulis, E. Barthelemy, R. Roemmich, C.J. Hass, Postural instability and gait impairment during obstacle crossing in parkinson's disease, Arch. Phys. Med. Rehabil. 93 (2012) 703–709. doi:10.1016/j.apmr.2011.11.004.

[24] M. Arvin, M. Mazaheri, M.J.M. Hoozemans, M. Pijnappels, B.J. Burger, S.M.P. Verschueren, J.H. van Dieën, Effects of narrow base gait on mediolateral balance control in young and older adults, J. Biomech. 49 (2016) 1264–1267. doi:10.1016/j.jbiomech.2016.03.011.

[25] C.J. Hass, D.E. Waddell, R.P. Fleming, J.L. Juncos, R.J. Gregor, Gait initiation and

dynamic balance control in Parkinson’s disease, Arch. Phys. Med. Rehabil. 86 (2005) 2172–2176. doi:10.1016/j.apmr.2005.05.013.

[26] H.A. Chang, D.E. Krebs, Dynamic balance control in elders: Gait initiation assessment as a screening tool, Arch. Phys. Med. Rehabil. 80 (1999) 490–494. doi:10.1016/S0003-9993(99)90187-9.

[27] M. Martin, M. Shinberg, M. Kuchibhatla, L. Ray, J.J. Carollo, M.L. Schenkman, Gait Initiation in Community-Dwelling Adults With Parkinson Disease: Comparison With Older and Younger Adults Without the Disease, Phys. Ther. 82 (2002) 566–577. doi:10.1093/ptj/82.6.566.

[28] J. Mickelborough, M.L. Van Der Linden, R.C. Tallis, A.R. Ennos, Muscle activity during gait initiation in normal elderly people, Gait Posture. 19 (2004) 50–57. doi:10.1016/S0966-6362(03)00016-X.

[29] H.A. Bischoff-Ferrari, M. Conzelmann, H.B. Stähelin, W. Dick, M.G. Carpenter, A.L. Adkin, R. Theiler, M. Pfeifer, J.H.J. Allum, Is fall prevention by vitamin D mediated by a change in postural or dynamic balance?, Osteoporos. Int. 17 (2006) 656–663. doi:10.1007/s00198-005-0030-9.

[30] J. Gill, J.H.J. Allum, M.G. Carpenter, M. Held-Ziolkowska, A.L. Adkin, F. Honegger, K. Pierchala, Trunk sway measures of postural stability during clinical balance tests: Effects of age, Journals Gerontol. - Ser. A Biol. Sci. Med. Sci. 56

(2001) 438–447. doi:10.1093/gerona/56.7.M438.

[31] L.D. RA, Mann, Hagy JL, White V, The initiation of gait, J. Bone Jt. Surgeryery. 61 (1979) 232–9.

[32] L. Laudani, A. Casabona, V. Perciavalle, A. Macaluso, Control of head stability during gait initiation in young and older women, J. Electromyogr. Kinesiol. (2006). doi:10.1016/j.jelekin.2006.08.001.

[33] R. Lepers, Y. Brenière, The role of anticipatory postural adjustments and gravity in gait initiation, Exp. Brain Res. 107 (1995) 118–124. doi:10.1007/BF00228023.

[34] J.C. Ceccato, M. de Sèze, C. Azevedo, J.R. Cazalets, Comparison of trunk activity during gait initiation and walking in humans, PLoS One. 4 (2009). doi:10.1371/journal.pone.0008193.

[35] C.P. Hurt, N. Rosenblatt, J.R. Crenshaw, M.D. Grabiner, Variation in trunk kinematics influences variation in step width during treadmill walking by older and younger adults, Gait Posture. (2010). doi:10.1016/j.gaitpost.2010.02.001.

[36] M.D. Grabiner, S. Donovan, M. Lou Bareither, J.R. Marone, K. Hamstra-Wright, S. Gatts, K.L. Troy, Trunk kinematics and fall risk of older adults: Translating biomechanical results to the clinic, J. Electromyogr. Kinesiol. 18 (2008) 197–204. doi:10.1016/j.jelekin.2007.06.009.

[37] S.W. Lee, J. Verghese, R. Holtzer, J.R. Mahoney, M. Oh-Park, Trunk sway during walking among older adults: Norms and correlation with gait velocity, Gait Posture. 40 (2014) 676–681. doi:10.1016/j.gaitpost.2014.07.023.

[38] N. Shishov, Y. Gimmon, H. Rashed, I. Kurz, R. Riemer, A. Shapiro, R. Debi, I. Melzer, Old adult fallers display reduced flexibility of arm and trunk movements when challenged with different walking speeds, Gait Posture. 52 (2017) 280–286. doi:10.1016/j.gaitpost.2016.12.004.

[39] M.J. Pavol, T.M. Owings, K.T. Foley, M.D. Grabiner, Gait characteristics as risk factors for falling from trips induced in older adults, Journals Gerontol. - Ser. A Biol. Sci. Med. Sci. 54 (1999) 583–590. doi:10.1093/gerona/54.11.M583.

[40] C.R. Lowrey, R.J. Reed, L.A. Vallis, Control strategies used by older adults during multiple obstacle avoidance, Gait Posture. 25 (2007) 502–508. doi:10.1016/j.gaitpost.2006.05.012.

[41] M.E. Hahn, L.S. Chou, Age-related reduction in sagittal plane center of mass motion during obstacle crossing, J. Biomech. 37 (2004) 837–844. doi:10.1016/j.jbiomech.2003.11.010.

[42] M. Fujimoto, L.S. Chou, Sagittal plane momentum control during walking in elderly fallers, Gait Posture. 45 (2016) 121–126. doi:10.1016/j.gaitpost.2016.01.009.

[43] M. Vlutters, E.H.F. Van Asseldonk, H. Van Der Kooij, Center of mass velocity-

based predictions in balance recovery following pelvis perturbations during human walking, J. Exp. Biol. 219 (2016) 1514–1523. doi:10.1242/jeb.129338.

[44] J.L. Honeine, M. Schieppati, O. Gagey, M.C. Do, The Functional Role of the Triceps Surae Muscle during Human Locomotion, PLoS One. 8 (2013). doi:10.1371/journal.pone.0052943.

[45] N. Chastan, G.W.M. Westby, S.T. du Montcel, M.C. Do, R.K. Chong, Y. Agid, M.L. Welter, Influence of sensory inputs and motor demands on the control of the centre of mass velocity during gait initiation in humans, Neurosci. Lett. 469 (2010) 400–404. doi:10.1016/j.neulet.2009.12.038.

[46] M.F. Vieira, A.A. de Brito, G.C. Lehnen, F.B. Rodrigues, Center of pressure and center of mass behavior during gait initiation on inclined surfaces: A statistical parametric mapping analysis, J. Biomech. (2017). doi:10.1016/j.jbiomech.2017.02.018.

[47] J.C. Singer, S.D. Prentice, W.E. McIlroy, Age-related changes in mediolateral dynamic stability control during volitional stepping, Gait Posture. 38 (2013) 679–683. doi:10.1016/j.gaitpost.2013.03.003.

[48] S.E. Halliday, D.A. Winter, J.S. Frank, A.E. Patla, N.L.G. Ontario, halliday_GaitPosture_1998, 8 (1998) 1–7. doi:10.1016/S0966-6362(98)00020-4.

[49] K.M. Means, The obstacle course: A tool for the assessment of functional balance

and mobility in the elderly, J. Rehabil. Res. Dev. 33 (1996) 413–428.

[50] L.Z. Rubenstein, K.R. Josephson, P.R. Trueblood, K. Yeung, J.O. Harker, A.S. Robbins, The reliability and validity of an obstacle course as a measure of gait and balance in older adults, Aging Clin. Exp. Res. 9 (1997) 127–135. doi:10.1007/bf03340138.

[51] D. Brunt, V. Santos, H.D. Kim, K. Light, C. Levy, Initiation of movement from quiet stance: Comparison of gait and stepping in elderly subjects of different levels of functional ability, Gait Posture. 21 (2005) 297–302. doi:10.1016/j.gaitpost.2004.03.003.

[52] V. Lugade, V. Lin, L.S. Chou, Center of mass and base of support interaction during gait, Gait Posture. 33 (2011) 406–411. doi:10.1016/j.gaitpost.2010.12.013.

[53] M.C. Nevitt, S.R. Cummings, E.S. Hudes, Risk factors for injurious falls: A prospective study, Journals Gerontol. 46 (1991) 164–170. doi:10.1093/geronj/46.5.M164.

[54] S. Porter, J. Nantel, Older adults prioritize postural stability in the anterior-posterior direction to regain balance following volitional lateral step, Gait Posture. 41 (2015) 666–669. doi:10.1016/j.gaitpost.2015.01.021.

[55] P.J. Sparto, J.R. Jennings, J.M. Furman, M.S. Redfern, Lateral step initiation behavior in older adults, Gait Posture. 39 (2014) 799–803.

doi:10.1016/j.gaitpost.2013.10.021.

[56] M.A. Townsend, Biped gait stabilization via foot placement, J. Biomech. (1985). doi:10.1016/0021-9290(85)90042-9.

[57] T.M. Owings, M.D. Grabiner, Variability of step kinematics in young and older adults, Gait Posture. 20 (2004) 26–29. doi:10.1016/S0966-6362(03)00088-2.

[58] C.E. Bauby, A.D. Kuo, Active control of lateral balance in human walking, J. Biomech. 33 (2000) 1433–1440. doi:10.1016/S0021-9290(00)00101-9.

[59] M. ML, R.M. , Johnson ME, Martinez KM, Age-dependent differences in lateral balance recovery through protective stepping, Clin. Biomech. 20 (2005) 607–16.

[60] S.C. Tseng, S.J. Stanhope, S.M. Morton, Impaired reactive stepping adjustments in older adults, Journals Gerontol. - Ser. A Biol. Sci. Med. Sci. 64 (2009) 807–815. doi:10.1093/gerona/glp027.

[61] T. Caderby, E. Yiou, N. Peyrot, M. Begon, G. Dalleau, Influence of gait speed on the control of mediolateral dynamic stability during gait initiation, J. Biomech. (2014). doi:10.1016/j.jbiomech.2013.11.011.

[62] B.E. Maki, P.J. Holliday, A.K. Topper, A prospective study of postural balance and risk of falling in an ambulatory and independent elderly population, Journals Gerontol. 49 (1994) 72–84. doi:10.1093/geronj/49.2.M72.

[63] T. Ito, T. Azuma, N. Yamashita, Anticipatory control in the initiation of a single step under biomechanical constraints in humans, Neurosci. Lett. 352 (2003) 207–210. doi:10.1016/j.neulet.2003.09.002.

[64] R. Herman, T. Cook, B. Cozzens, W. Freedman, Control of posture and locomotion, second, New York: Plenum Press, New York, NY, 1973.

[65] Y. Brenière, M.C. Do, J. Sanchez, A biomechanical study of the gait initiation process, J. Biophys. Med. Nucl. 5 (1981) 197–205.

[66] D.A. Winter, Human balance and posture control during standing and walking, Gait Posture. 3 (1995) 193–214. doi:10.1016/0966-6362(96)82849-9.

[67] C. Hsiang Tang, K. D E, Dynamic balance control in elders: gait initiation assessment as a screening tool., Arch. Phys. Med. Rehabil. 80 (1999) 490–4. doi:10.1016/s0003-9993(99)90187-9.

[68] J.L. Zettel, W.E. McIlroy, B.E. Maki, Environmental constraints on foot trajectory reveal the capacity for modulation of anticipatory postural adjustments during rapid triggered stepping reactions, Exp. Brain Res. 146 (2002) 38–47. doi:10.1007/s00221-002-1150-5.

[69] H.C. Diener, F.B. Horak, L.M. Nashner, Influence of central set on human postural responses, J. Neurophysiol. 59 (1988) 1888–1905. doi:10.1152/jn.1989.62.4.841.

[70] T. Onushko, T. Boerger, J. Van Dehy, B.D. Schmit, Dynamic stability and stepping strategies of young healthy adults walking on an oscillating treadmill, PLoS One. 14 (2019) 1–12. doi:10.1371/journal.pone.0212207.

[71] S. Leteneur, E. Simoneau, C. Gillet, Y. Dessery, F. Barbier, Trunk's Natural Inclination Influences Stance Limb Kinetics, but Not Body Kinematics, during Gait Initiation in Able Men, PLoS One. 8 (2013). doi:10.1371/journal.pone.0055256.

[72] D.A. Winter, Biomechanics and Motor Control of Human Movement, fourth, John Wiley & Sons, Inc, Waterloo, Ontario, 2009.

[73] C.D. MacKinnon, D.A. Winter, Control of whole body balance in the frontal plane during human walking, J. Biomech. (1993). doi:10.1016/0021-9290(93)90027-C.

[74] M. Arvin, J.H. van Dieën, S.M. Bruijn, Effects of constrained trunk movement on frontal plane gait kinematics, J. Biomech. (2016). doi:10.1016/j.jbiomech.2016.07.015.

[75] K.M. Newell, D.E. Vaillancourt, J.J. Sosnoff, Aging, Complexity, and Motor Performance, in: J.E. Birren, K.W. Schaie, R.P. Abeles, M. Gatz, T.A. Salthouse (Eds.), Handb. Psychol. Aging, sixth, Elsevier Inc., Amsterdam, 2006: pp. 163–182. doi:10.1016/B978-012101264-9/50011-2.

[76] J.M. Hausdorff, D.A. Rios, H.K. Edelberg, Gait variability and fall risk in community-living older adults: A 1-year prospective study, Arch. Phys. Med.

Rehabil. (2001). doi:10.1053/apmr.2001.24893.

[77] K. Saladin, Anatomy and Physiology: The Unity of Form and Function, New York: McGraw Hill, 2012.

[78] R.T. Harbourne, N. Stergiou, Movement Variability and the Use of Nonlinear Tools: Principles to Guide Physical Therapist Practice, Phys. Ther. 89 (2009) 267–282. doi:10.2522/ptj.20080130.

[79] N. Cau, V. Cimolin, M. Galli, H. Precilios, E. Tacchini, C. Santovito, P. Capodaglio, Center of pressure displacements during gait initiation in individuals with obesity, J. Neuroeng. Rehabil. 11 (2014) 1–8. doi:10.1186/1743-0003-11-82.

[80] H.D. Kim, D. Brunt, The Effect of a Dual-Task on Obstacle Crossing in Healthy Elderly and Young Adults, Arch. Phys. Med. Rehabil. 88 (2007) 1309–1313. doi:10.1016/j.apmr.2007.07.001.

[81] C.J. Hass, D.E. Waddell, S.L. Wolf, J.L. Juncos, R.J. Gregor, Gait initiation in older adults with postural instability, Clin. Biomech. (2008). doi:10.1016/j.clinbiomech.2008.02.012.

[82] M. Hoffman, J. Schrader, T. Applegate, D. Koceja, Unilateral postural control of the functionally dominant and nondominant extremities of healthy subjects, J. Athl. Train. 33 (1998) 319–322.

[83] H.F. Pan, H.C. Hsu, W.N. Chang, J.H. Renn, H.W. Wu, Strategies for obstacle crossing in older adults with high and low risk of falling, J. Phys. Ther. Sci. 28 (2016) 1614–1620. doi:10.1589/jpts.28.1614.

[84] A.E. Patla, Visual control of human locomotion, Adv. Psychol. 78 (1991) 55–97. doi:10.1007/978-3-030-31866-6_75.

[85] D.A. Winter, A.B.C. (anatomy, biomechanics and control) of balance during standing and walking, Waterloo, Biomechanics, 1995.

[86] R.L. Wright, D.M. Peters, P.D. Robinson, T.N. Watt, M.A. Hollands, Older adults who have previously fallen due to a trip walk differently than those who have fallen due to a slip, Gait Posture. 41 (2015) 164–169. doi:10.1016/j.gaitpost.2014.09.025.

[87] H.J. Lee, L.S. Chou, Detection of gait instability using the center of mass and center of pressure inclination angles, Arch. Phys. Med. Rehabil. 87 (2006) 569–575. doi:10.1016/j.apmr.2005.11.033.

[88] Y.C. Pai, J. Patton, Center of mass velocity-position predictions for balance control, J. Biomech. 30 (1997) 347–354. doi:10.1016/S0021-9290(96)00165-0.

[89] C.J. Hasson, R.E.A. Van Emmerik, G.E. Caldwell, Predicting dynamic postural instability using center of mass time-to-contact information, J. Biomech. 41 (2008) 2121–2129. doi:10.1016/j.jbiomech.2008.04.031.

[90] J.C. Singer, S.D. Prentice, W.E. McIlroy, Age-related changes in mediolateral dynamic stability control during volitional stepping, Gait Posture. 38 (2013) 679–683. doi:10.1016/j.gaitpost.2013.03.003.

[91] J.-L. Honeine, M. Schieppati, O. Gagey, M.-C. Do, By counteracting gravity, triceps surae sets both kinematics and kinetics of gait, Physiol. Rep. 2 (2014) e00229. doi:10.1002/phy2.229.

[92] K. Uemura, M. Yamada, K. Nagai, M. Shinya, N. Ichihashi, Effect of dual-tasking on the center of pressure trajectory at gait initiation in elderly fallers and non-fallers, Aging Clin. Exp. Res. 24 (2012) 152–156. doi:10.1007/BF03325161.

[93] F.G. Maki BE, Holliday PJ, Aging and postural control. A comparison of spontaneous- and induced-sway balance tests, J. Am. Geriatr. Soc. 38 (1990) 1–9.

[94] M.H. Morcelli, D.P. LaRoche, L.F. Crozara, N.R. Marques, C.Z. Hallal, D.M. Rossi, M. Gonçalves, M.T. Navega, Neuromuscular performance in the hip joint of elderly fallers and non-fallers, Aging Clin. Exp. Res. 28 (2016) 443–450. doi:10.1007/s40520-015-0448-7.

[95] N.R. Marques, D.P. Laroche, C.Z. Hallal, L.F. Crozara, M.H. Morcelli, A.H. Karuka, M.T. Navega, M. Gonçalves, Association between energy cost of walking, muscle activation, and biomechanical parameters in older female fallers and non-fallers, Clin. Biomech. 28 (2013) 330–336. doi:10.1016/j.clinbiomech.2013.01.004.

[96] C.A. McGibbon, M.S. Puniello, D.E. Krebs, Mechanical energy transfer during gait in relation to strength impairment and pathology in elderly women, Clin. Biomech. 16 (2001) 324–333. doi:10.1016/S0268-0033(01)00004-3.

[97] S. Sasaki, Y. Nagano, S. Kaneko, S. Imamura, T. Koabayshi, T. Fukubayashi, The relationships between the center of mass position and the trunk, hip, and knee kinematics in the sagittal plane: A pilot study on field-based video analysis for female soccer players, J. Hum. Kinet. 45 (2015) 71–80. doi:10.1515/hukin-2015-0008.

[98] B.E. Maki, Gait changes in older adults: Predictors of falls or indicators of fear?, J. Am. Geriatr. Soc. 45 (1997) 313–320. doi:10.1111/j.1532-5415.1997.tb00946.x.

[99] S.R. Lord, M.W. Rogers, A. Howland, R. Fitzpatrick, Lateral stability, sensorimotor function and falls in older people, J. Am. Geriatr. Soc. 47 (1999) 1077–1081. doi:10.1111/j.1532-5415.1999.tb05230.x.

[100] J. Panula, H. Pihlajamäki, V.M. Mattila, P. Jaatinen, T. Vahlberg, P. Aarnio, S.L. Kivelä, Mortality and cause of death in hip fracture patients aged 65 or older - A population-based study, BMC Musculoskelet. Disord. 12 (2011) 105. doi:10.1186/1471-2474-12-105.

[101] E. Yiou, T. Caderby, A. Delafontaine, P. Fourcade, J.-L. Honeine, Balance control during gait initiation: State-of-the-art and research perspectives, World J. Orthop. 8

(2017) 815–828. doi:10.5312/wjo.v8.i11.815.

[102] T. Caderby, E. Yiou, N. Peyrot, M. Begon, G. Dalleau, Influence of gait speed on the control of mediolateral dynamic stability during gait initiation, J. Biomech. 47 (2014) 417–423. doi:10.1016/j.jbiomech.2013.11.011.

[103] E. Yiou, P. Fourcade, R. Artico, T. Caderby, Influence of temporal pressure constraint on the biomechanical organization of gait initiation made with or without an obstacle to clear, Exp. Brain Res. 234 (2016) 1363–1375. doi:10.1007/s00221-015-4319-4.

[104] B.E. Maki, W.E. McIlroy, Control of compensatory stepping reactions: age-related impairment and the potential for remedial intervention, PHYSIOTHER THEORY Pr. 15 (n.d.) 69–90.

[105] I.N. Lyon, B.L. Day, Control of frontal plane body motio in human stepping, Exp Brain Res 115 (1997) 345-356.

[106] J.L. Zettel, W.E. McIlroy, B.E. Maki, Can stabilizing features of rapid triggered stepping reactions be modulated to meet environmental constraints?, Exp. Brain Res. 145 (2002) 297–308. doi:10.1007/s00221-002-1083-z.

[107] M.A. Schrager, V.E. Kelly, R. Price, L. Ferrucci, A. Shumway-Cook, The effects of age on medio-lateral stability during normal and narrow base walking, Gait Posture. 28 (2008) 466–471. doi:10.1016/j.gaitpost.2008.02.009.

[108] B.E. Maki, E. MA, W.E. McIlroy, Age-related differences in laterally directed compensatory stepping behavior, J. Gerontol. Biol. Sci. 55 (2000) 270–277.

[109] L.S. Chou, K.R. Kaufman, M.E. Hahn, R.H. Brey, Medio-lateral motion of the center of mass during obstacle crossing distinguishes elderly individuals with imbalance, Gait Posture. 18 (2003) 125–133. doi:10.1016/S0966-6362(02)00067-X.

[110] C.R. Lowrey, A. Watson, L.A. Vallis, Age-related changes in avoidance strategies when negotiating single and multiple obstacles, Exp. Brain Res. 182 (2007) 289–299. doi:10.1007/s00221-007-0986-0.

[111] M.W. Rogers, M.L. Mille, Lateral stability and falls in older people, Exerc. Sport Sci. Rev. 31 (2003) 182–187. doi:10.1097/00003677-200310000-00005.

[112] J.W. Park, M. Jung, M. Kweon, The mediolateral CoP parameters can differentiate the fallers among the community-dwelling elderly population, J. Phys. Ther. Sci. 26 (2014) 381–384. doi:10.1589/jpts.26.381.

[113] B.E. Maki, P.J. Holliday, A.K. Topper, Fear of falling and postural performance in the elderly, Journals Gerontol. 46 (1991). doi:10.1093/geronj/46.4.M123.

[114] M. Schenkman, K.M. Shipp, J. Chandler, S.A. Studenski, M. Kuchibhatla, Relationships between mobility of axial structures and physical performance, Phys. Ther. 76 (1996) 276–285. doi:10.1093/ptj/76.3.276.

[115] N.B. Alexander, Postural Control in Older Adults, J. Am. Geriatr. Soc. 42 (1994) 93–108. doi:10.1111/j.1532-5415.1994.tb06081.x.

[116] S.R. Lord, D.L. Sturnieks, The physiology of falling: Assessment and prevention strategies for older people, J. Sci. Med. Sport. 8 (2005) 35–42. doi:10.1016/S1440-2440(05)80022-2.

[117] Y. Lajoie, S.P. Gallagher, Predicting falls within the elderly community: Comparison of postural sway, reaction time, the Berg balance scale and the Activities-specific Balance Confidence (ABC) scale for comparing fallers and non-fallers, Arch. Gerontol. Geriatr. 38 (2004) 11–26. doi:10.1016/S0167-4943(03)00082-7.

[118] J.J. Craig, A.P. Bruetsch, J.M. Huisinga, Coordination of trunk and foot acceleration during gait is affected by walking velocity and fall history in elderly adults, Aging Clin. Exp. Res. 31 (2019) 943–950. doi:10.1007/s40520-018-1036-4.

[119] S.G. Brauer, Y.R. Burns, P. Galley, A prospective study of laboratory and clinical measures of postural stability to predict community-dwelling fallers, Journals Gerontol. - Ser. A Biol. Sci. Med. Sci. 55 (2000) 469–476. doi:10.1093/gerona/55.8.M469.

[120] M.J. Hilliard, K.M. Martinez, I. Janssen, M. M-l, Y. Zhang, R. Mw, Lateral Balance Factors Predict Future Falls in Community-Living Older Adults, 89 (2008) 1708–

1713. doi:10.1016/j.apmr.2008.01.023.

[121] P. Cesar, B. Bento, G. Pereira, C. Ugrinowitsch, A. Luiz, F. Rodacki, Peak torque and rate of torque development in elderly with and without fall history, Clin. Biomech. 25 (2010) 450–454. doi:10.1016/j.clinbiomech.2010.02.002.