FINAL PROGRESS REPORT

Effects of Obesity and Age on Fall Risk - Implications for Safety Guidelines

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ABSTRACT

Falls are a significant cause of occupational morbidity and mortality. The number of obese and obese older workers in the US is growing. Recent research indicates obese workers are at a higher risk of falls compared to non-obese workers, but the reasons for this higher risk are unclear. These reasons need to be better understood in order to develop strategies for fall prevention that are inclusive of the needs and characteristics of obese workers.

Based upon this need, the goal of this application was to more fully characterize the effects of obesity on balance and risk of falls. Our work focused on falls caused by slips and trips, which are among the most commonly reported contributing factors to occupational falls. We also investigated young and older adults given growing number of older workers, and the potential for differential effects of obesity among these two demographics. Our overarching hypothesis was that balance and risk of falls from slips, trips, and loss of balance are adversely influenced by obesity, age, and their interaction. Balance and fall risk were assessed by estimating the risk of losing balance (due to slipping or tripping) as well as the ability to recovery balance once it is lost (after slipping or tripping). Our work investigated both components of fall risk (losing balance and recovering balance once it is lost) since they are distinct processes and may be differentially affected by obesity and age. Such differential effects would in turn suggest different intervention approaches.

Our first aim was to *determine the effects of age, obesity, and their interaction on the risk of slipping, tripping, and loss of balance during simulated construction work*. Obesity did not increase the risk of slipping while walking among young and older adults, including during occupationally-relevant tasks of ramp walking and load carriage. Obesity did not increase the risk of tripping among young adults, but did increase the risk of tripping among older adults. Age did not increase the risk of slipping or tripping. Our second aim was to *determine the effects of age, obesity, and their interaction on balance and balance recovery after slipping and tripping.* Obesity increased the rate of falling after exposure to a laboratory-induced slip and trip, suggesting obesity adversely affects the ability to recover balance after it is lost. Age did not adversely affect the rate of falling after slipping. Obesity was also associated with a deficient compensatory stepping response after slipping and tripping, and lower strength relative to body mass at the ankle, knee, and hip. Both the compensatory stepping response and strength are to aspects of balance and fall prevention that are amenable to intervention.

The results from this work clarify the reasons for the higher fall rate among obese individuals, and provide direction for subsequent work to develop fall prevention interventions. Data was also obtained that can be used to evaluate the expected effectiveness of established fall prevention safety guidelines among obese workers, and that that might be helpful in developing engineering/administrative controls have the potential to reduce the risk of falls.

SIGNIFICANT (KEY) FINDINGS

Our first aim was to *determine the effects of age, obesity, and their interaction on the risk of slipping, tripping, and loss of balance during simulated construction work.*

Risk of slipping: Obesity did not increase such risk while walking over level ground, descending a ramp, or during load carriage (Figure 1). Advanced age also did not increase the risk of slipping while walking. Prior work has shown that older adults are at a lower risk of slipping while walking. However, it was unclear if this was an inherent age effect, or because older adults tend to walk at a slower speed and with shorter step length. Our results indicated that older adults are at a lower risk of slipping when walking at a controlled speed without controlling step length. When both age groups walked with the same speed and step length, however, no age-related differences were found. Thus, gait speed and step length can have a substantial influence on slip risk, and the previously reported effects of age on risk of slipping appear to be due to differences in speed and step length rather than inherent age differences. Speed and step length were also found to have independent and opposite effects on the risk of slipping, with step length having a strong positive effect, and speed a weaker negative effect. *These results are important in indicating that the reported higher fall rate among obese or older individuals is not likely due to slipping more frequently.*

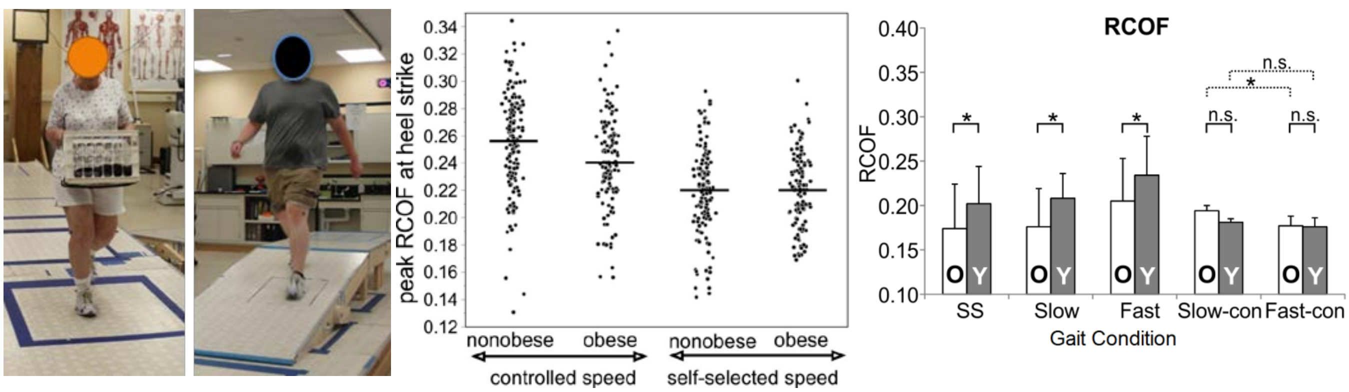

Figure 1. Left pane: Experimental protocol for determining the effects of obesity and age on the risks of slipping and tripping during level walking, ramp descent, load carriage. Middle pane: Results showing no effect of obesity (*p*>0.05) on risk of slipping while walking at a self-selected speed and at a hurried controlled speed. Risk of slipping was quantified using the required coefficient of friction (RCOF), with a higher RCOF indicating a higher risk of slipping. Right pane: Results showing a lower RCOF (and risk of slipping) among older participants at self-selected (SS), slow, and fast gait speeds without controlling step length (Slow and Fast), but no difference in RCOF when controlling step length (Slow-con and Fast-con).

Risk of tripping: We developed two improved analytical methods to assess the risk of tripping while walking as a function of potential obstacle height, and to compare the risk of tripping between participant groups. Using these new methods, we found a higher risk of tripping among obese older adults compared to healthy-weight older adults for obstacle heights of 2.4 - 4.2 cm (Figure 2), but no effects of obesity on the risk of tripping among young adults. We also found a lower risk of tripping among older vs. young adults and a higher risk of tripping among females vs. males (not shown). Anterior load carriage did not increase the risk of tripping, but may obstruct the view of potential tripping obstacles. Among obese and non-obese participants, performing mental dual tasks while walking elicited changes in gait that increased tripping risk. *These results provide evidence for an adverse effect of obesity on the risk of tripping among older, but not young, adults.*

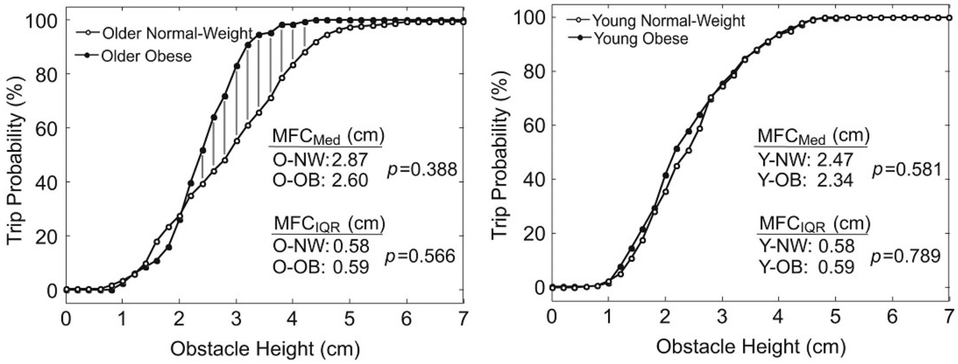

Figure 2. Results illustrating the effects of obesity among older adults (left pane) and young adults (right pane) on risk of tripping as quantified by trip probability when encountering virtual obstacles ranging in height from 0 to 7 cm. Vertical lines indicate significant differences between participant groups at a particular obstacle height.

Our second aim was to *determine the effects of age, obesity, and their interaction on balance and balance recovery after slipping and tripping.*

Balance recovery after slipping: After a laboratory-induced slip, obese participants fell at a rate of 32% compared to 10% among non-obese participants (Figure 3). A logistic regression model indicated obese participants were 8x more likely to fall after slipping compared to non-obese individuals. The slips experienced by obese participants were also 22% faster than those among non-obese individuals. No age effects were found for fall rate or slip speed. *These results suggest that the higher fall rate among obese individuals likely results, at least in part, from a* ***greater difficulty recovering balance*** *after slipping.* Slip distance thresholds were also identified that may have value in developing controls for fall prevention.

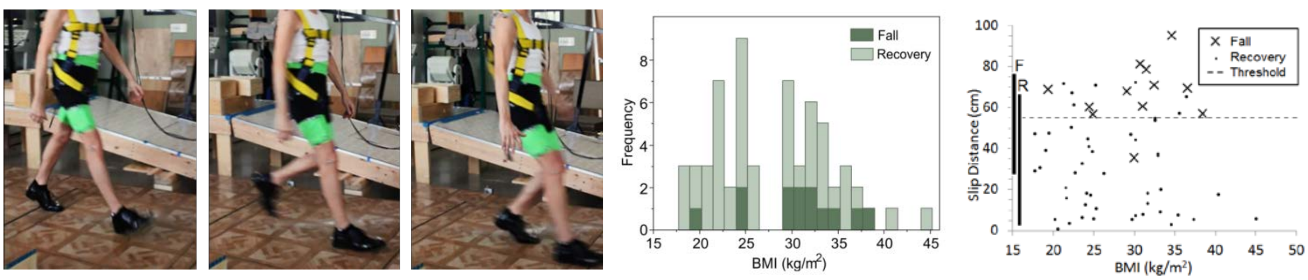

Figure 3. Left pane: Participant being exposed to a laboratory-induced slip using vegetable oil. Middle pane: Histogram illustrating the number of falls and recoveries as a function of body mass index. Right pane: Slip distance during falls and recoveries as a function of body mass index. The dashed horizontal line represents a slip distance of 56.5 cm, which separated 85.4% of recoveries from 86.7% of falls. Thus, limiting slip distance to below this value though engineering controls such as floor markings or texture may help to reduce the number of slip-induced falls.

Balance recovery after tripping: After a laboratory-induced trip, obese participants fell at a rate of 30% compared to 0% among non-obese participants (Figure 4). Falls were more often associated with a "lowering strategy" to recover balance when stepping (tripped foot lowered to floor and contralateral foot stepped over obstacle), a longer time to complete the initial balance recovery step, a larger trunk angle and angular velocity during recovery, and a lower hip height during recovery. Similarly, obesity was associated with using a lowering strategy and a longer time to complete the initial recovery step. *These results indicate that the higher fall rate among* *obese individuals likely results, at least in part, from a* ***greater difficulty recovering balance*** *after tripping, and that this greater difficulty is due to a deficient stepping response.* As such, fall prevention interventions that focus on improving the stepping response should help to reduce the fall rate among obese individuals.

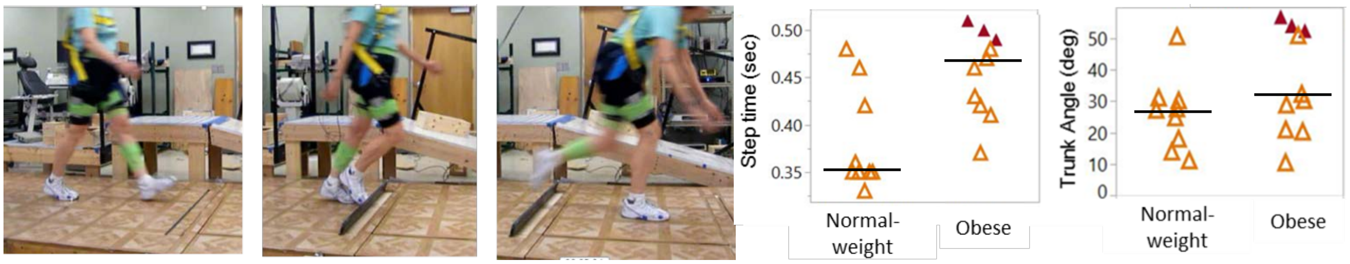

Figure 4. Left pane: Participant being exposed to a laboratory trip using a hidden obstacle that suddenly raises 7 cm. Middle pane: Median step time was higher among obese participants, and higher among falls (solid triangles). Right pane: Median trunk angle during recovery was not affected by obesity.

We also completed the first comprehensive evaluation of the influence of obesity on lower extremity strength. Obesity-related differences in lower extremity strength were mostly consistent between young and older adults, with only one significant obesity x age interaction effect observed (hip flexor strength). However, the influence of obesity substantially differed between joints/exertion combinations. Moreover, obese adults used higher joint moments relative to available strength at the knee and ankle during gait, which could explain gait limitations among the obese. *Since lower extremity strength is a critical component of gait, balance, and fall prevention, these results illustrate the important effects of obesity that need to be addressed through intervention to improve balance recovery and fall rates.*

We also completed ancillary studies investigating sensory systems involved in balance control. We showed that obesity adversely affects plantar sensitivity, and that this adverse effect is associated with poorer standing balance exhibited by obese individuals. *These results provide new information on a physiological mechanism by which obesity adversely affects balance (reduced plantar sensitivity).*

TRANSLATION OF FINDINGS

The findings of this project cannot yet be applied to the workplace. However, the findings provide important direction for future efforts aimed at reducing falls among obese workers.

- Our findings indicate that the reported higher fall rate among obese and/or older individuals/workers is not likely due to slipping more frequently.

- Our findings provide evidence for an adverse effect of obesity on the risk of tripping among older, but not young, workers. As such, older obese workers may experience more frequent trips than older non-obese workers.

- Our findings suggest that the higher fall rate among obese individuals/workers is likely due, at least in part, to a greater difficulty recovering balance after slipping or tripping. As such, interventions aimed at improving the compensatory stepping response after these large postural perturbations should help to reduce the fall rate among obese workers.

- The slip distance that separated falls from recoveries was determined. Engineering controls, such as adding floor textures that limit slip distance below this threshold if a slip was to occur, may help to reduce slip-induced falls. This idea requires laboratory validation.

- We developed a new method of visualizing the risk of slipping (and tripping) as a function of floor/shoe friction characteristics (and gait patterns). This method can be used to visualize the potential effectiveness of safety guidelines, as well as to determine whether these safety guidelines need to be modified to be inclusive of all workers.

OUTCOME/IMPACT

Potential outcomes

- The findings of this research provide important guidance for future investigations aimed at developing fall prevention strategies for obese workers. We determined that obese individuals/workers are, in general, not at a greater risk of losing balance due to slipping or tripping while walking, but that they have greater difficulty recovering balance after slipping or tripping. These findings suggest fall prevention efforts for obese workers should focus on improving the ability to recover balance after slipping or tripping. We have on-going efforts in this regard.
- We have developed improved research methods to evaluate the influence of obesity (and other factors) on risk of slipping and tripping. These methods can also be used to indirectly assess the efficacy of safety guidelines related to preventing slipping and tripping, and determine if any such guidelines will be inclusive with the needs of different worker populations.
- We have also established slip distance thresholds that could be used for the development of engineering controls/floor designs to prevent slip-induced falls. These thresholds need validation prior to being evaluated outside of the laboratory.

SCIENTIFIC REPORT

BACKGROUND

Importance of the problem

Falls are the leading cause of work-related deaths in the US construction industry, and the second leading cause of nonfatal injuries in construction resulting in days away from work (CPWR, 2007). Two major demographic trends in the US population threaten to further exacerbate the problem of falls in the construction industry: the aging of the construction workforce (CPWR, 2007), and the increasing prevalence of obesity (CDC, 2006). Several studies have reported an increased likelihood of injury or death from a fall in older workers (Courtney et al., 2001; HMSO, 1985; Kisner et al., 1997; Loomis et al., 1997; Marsh et al., 2001), and a growing number of studies suggest that being overweight or obese also increases the risk for falls and fall-related injuries (Finkelstein et al., 2007; Fjeldstad et al., 2008). Workers who are both older and obese would seem to be at an even higher risk due to the compounding effects of age and obesity on balance, though this has yet to be demonstrated. Approximately 78% of construction workers aged 55 and older were either overweight or obese in 2005 (CPWR, 2007), so these workers comprise a significant portion of the construction workforce. Because of these demographic trends, it important to understand how age and obesity contribute to falls so that safety guidelines, interventions, or controls can be developed that are protective of older and/or obese workers.

How the proposed work will improve scientific knowledge

Little is known about how obesity and age contribute to falls in the construction industry. A growing number of studies report impaired balance in obese adults during quiet standing (Chiari et al., 2002; Hue et al., 2007; Maffiuletti et al., 2005; McGraw et al., 2000; Messier et al., 2000; Sartorio et al., 2001; Teasdale et al., 2007). However, most falls in construction do not occur during quiet standing, but rather are caused by slips, trips, and loss of balance (Courtney et al., 2002; Hsiao et al., 2001; Layne et al., 2004; Lipscomb et al., 2003; Lipscomb et al., 2006). The sequence of major events that lead to a fall are summarized in Figure 1.

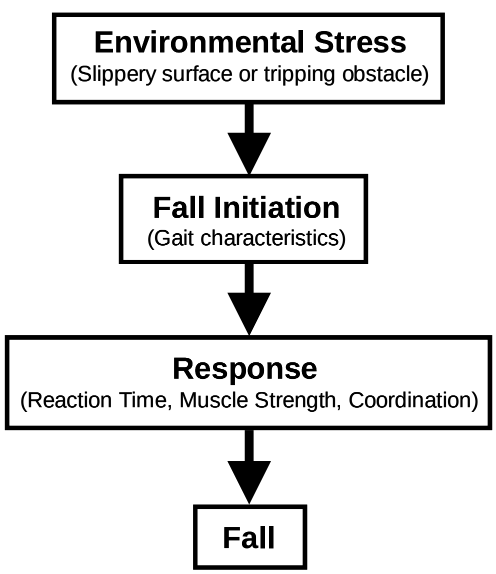

Figure 1. Conceptual model of the sequence of major events leading to a fall from a slip or trip. An environmental stress presents an increased risk for a slip or trip. A fall is initiated if gait characteristics are not protective (required coefficient of friction is high, or minimum ground clearance during the swing phase of gait is small). A postural response is then attempted upon detection of fall initiation, and is dependent upon neural and musculoskeletal function. If the biomechanical quality of the response fails to reach the demands of the task, then a fall will occur. It is unclear whether the increased number of occupational falls in older and/or obese individuals is due to altered gait characteristics increasing the number of falls initiated, or an impaired response once a fall has been initiated, or both. As such, the proposed work will investigate the effects of obesity, age, and their interaction on both.

Following these major events, the risk of falls is dependent upon 1) gait characteristics that modulate the number of falls initiated, and 2) the ability to respond to a fall initiation and recover balance without falling. The effects of obesity on gait characteristics that can affect risk of slipping/tripping are unclear. A previous study (Liu, 2010) and our preliminary data summarized below (Figure 4) indicate that overweight/obese individuals are at an increased risk of slipping, as evidenced by an increase in the required coefficient of friction to prevent a slip at the shoe/floor interface. Conversely, load carriage (which may be viewed as a surrogate to obesity since both result in an increase in effective body mass) decreases risk of slipping, as evidenced by a decrease in the required coefficient of friction (Cham et al., 2004). These conflicting reports highlight the need for the proposed work to clarify the effects of obesity on risk of slipping or tripping. Regarding the effects of obesity on the response to a slip, Leamon and Li (1991) reported an increase in slip severity (i.e. slip distance) with load carriage, which implies a poorer response to a slip initiating a fall. However, no studies have systematically investigated the effects of obesity and aging on fall initiation and responses.

Lateral buckling is an important source of loss of balance in construction. Lateral buckling is an instability condition caused when a worker is walking on a joist and a critical buckling load is exceeded. This results in lateral movement and twisting of an unbraced or partially braced joist or beam. Typically, the joists give no sign of buckling behavior until the critical buckling load is reached, and then the movement is dramatic. An estimated 2.4% of fall incidents result from lateral buckling (NIOSH, 2000). Unfortunately, there is currently little understanding of the loads or unbraced lengths at which lateral buckling will occur or the behavior of unbraced joists as workers walk or perform other construction tasks upon the joist.

The proposed work will improve our understanding of how age, obesity, and their combination contribute to falls in construction. The proposed work will investigate both fall initiation and recovery since they are distinct processes that may be differentially affected by age and obesity. Such differential effects would in turn suggest different intervention approaches. We acknowledge that we do not investigate the ability to detect a fall initiation, as this is less likely amenable to a workplace intervention. We will also focus on realistic construction activities to improve the external validity of our results.

Potential changes in practice if the proposed aims are achieved

Current fall prevention safety guidelines do not account for the characteristics of older and/or obese workers, or on balance demands during occupationally-relevant tasks. This project will provide the first biomechanical evidence of the effects of obesity and age on balance and risk of falls during both fundamental and relatively realistic construction tasks. These results can then be used by policy makers to develop more-inclusive safety guidelines and/or designs that account for the balance abilities of older and obese workers. For example, the current safety guideline for slip resistance is a coefficient of friction (COF) of 0.5. However, this was only recommended for constant speed walking over level surfaces, and was not based upon data from older and obese individuals. The results from the proposed work can be used to determine how age and obesity affect the COF not only for constant-speed walking on a level surface, but also for other occupationally-relevant tasks such as walking on inclined surfaces and load carriage. As another example, there are currently no safety guidelines with respect to lateral buckling on unbraced I-joists, which can lead to falls. Using our results, we envision an administrative control involving some form of ‘color-coding’ on the joists that indicates the portions of the beam that are susceptible to lateral buckling, and which account for effects of age and obesity on lateral buckling. This project addresses Strategic Goals *1.0 – Falls* and *12.0 – Disparities in Health and Safety in Construction* of the National Construction Agenda of

NORA. Though focused on construction-relevant activities, the results will also be applicable to many other occupational sectors.

INNOVATION

The application is innovative in three main ways. First, it is innovative in targeting the main and interactive effects of obesity, aging, and experience on balance in the construction industry. While the main effects of aging on balance have been studied extensively, the effects of obesity, experience, and their interaction with aging have received little attention despite the growing number of obese and obese older construction workers. Second, investigating these factors in realistic working conditions is innovative. For example, little is known about falls due to lateral buckling of joists. Only five studies on the lateral buckling of I-joists are known, three of which are authored by a member of our team (Hindman). Third, innovative analyses have been added to our approach to supplement the well-established measures proposed. These innovative analyses include using statistical modeling to estimate the probability of slipping and tripping during gait (Best et al., 2008; Chang, 2004; Chang et al., 2008), and using forward dynamic simulations to predict center-of-mass stability boundaries after loss of balance (Yang et al., 2007, 2008). These analyses can provide additional information to further our understanding of the effects of obesity, aging, and experience on balance and falls.

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SPECIFIC AIMS

Specific Aim #1: Determine the effects of age, obesity, and their interaction on the risk of slipping, tripping, and loss of balance during simulated construction work. This simulated work will involve: walking on level and inclined surfaces with and without load carriage (when a slip can occur); excessive obstacle height that poses a tripping hazard while walking on level and inclined surfaces (which can lead to a trip); and lateral buckling while walking along unbraced floor joists (which can lead to a loss of balance). Input from subject matter experts in residential construction firms will be used to inform the experimental design and ensure the realism of the simulated construction work. Both novices and actual construction workers will be recruited to investigate the effects of age and obesity on balance and balance recovery in each group, and to understand the how experience with construction work affects balance and balance recovery.

Specific Aim #2: Determine the effects of age, obesity, and their interaction on balance and balance recovery after slipping and tripping. In contrast to Specific Aim 1, which focuses on situations that can lead to a slip, trip, or loss of balance, this aim will focus on balance and balance recovery after losing balance. As in Specific Aim 1, both novice and actual construction workers will be recruited.

Age differences in the required coefficient of friction during level walking do not exist when experimentally-controlling speed and step length

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Abstract

The effects of gait speed and step length on the required coefficient of friction (COF) confounds the investigation of age-related differences in required COF. The goals of this study were to investigate whether age differences in required COF during self-selected gait persist when experimentally-controlling speed and step length, and to determine the independent effects of speed and step length on required COF. Ten young and ten older healthy adults performed gait trials under five gait conditions: self-selected, slow and fast speeds without controlling step length, and slow and fast speeds while controlling step length. During self-selected gait, older adults walked with shorter step lengths and exhibited a lower required COF. Older adults also exhibited a lower required COF when walking at a controlled speed without controlling step length. When both age groups walked with the same speed and step length, no age difference in required COF was found. Thus, speed and step length can have a large influence on studies investigating age-related differences in required COF. It was also found that speed and step length have independent and opposite effects on required COF, with step length having a strong positive effect on required COF, and speed a weaker negative effect.

Introduction

Falls are a major cause of injury and death among older adults. About 40% of community-dwelling adults age 65 and older fall each year, and the incidence of falls rises as age increases.[1] In addition, rates of injury and death related to falls increase with age[1,2] such that three quarters of deaths due to falls occur in people age 65 and over.[1] Slipping is the second most common cause of falls among older adults in cases of fall-related injury, accounting for about 21% of cases.[3] Thus, understanding the factors that contribute to slipping among older adults is important in the prevention of these falls.

The required coefficient of friction (COF), or utilized coefficient of friction, quantifies the minimum static friction necessary to prevent the foot from slipping,[4,5] and is calculated as the ratio of shear to vertical components of the ground reaction force (GRF). Understanding age-related changes in required COF may be important in reducing the incidence of slip-related falls among older adults. However, the nature of age effects on required COF remains uncertain. One study of walking on level surfaces reported lower required COF among older adults compared to young adults,[6] while others have reported no differences between older and young adults.[4,7-9] However, older adults tended to walk at a slower speed[6-9] and/or step length[8,9] compared to young adults, which could confound the identification of age differences in required COF.

Gait speed and step length influence the required COF and the shear and vertical components of the GRF. For example, required COF increases with increased step length,[4,10] and some authors have noted an expectation that gait speed should influence required COF as well.[5,8] Powers et al.[11] showed required COF increasing with increased speed, although increased speed was also accompanied by increases in step length. Thus, the effect of speed on required COF independent of step length remains unknown.

Based upon the incomplete understanding of how age, gait speed, and step length affect required COF, this study had two goals. The first goal was to investigate whether age differences in required COF during self-selected gait persist when experimentally-controlling speed and step length. The second goal was to determine the independent effects of speed and step length on required COF. Accomplishing these goals will provide fundamental information on how age, gait speed and step length affect required COF. It was hypothesized that 1) age differences in required COF would not persist when controlling both speed and step length, 2) increasing speed while holding step length constant would increase required COF, and 3) increasing step length while holding speed constant would increase required COF.

Methods

Twenty healthy adults participated including ten young adults (mean±standard deviation: age = 23.9±3.3 years, mass = 61.7±7.3 kg, height = 1.65±0.09 m) and ten older adults (mean age = 80.3±4.0 years, mass = 65.2±10.5 kg, height = 1.63±0.08 m). There were no differences between groups in mass (p = 0.396) or height (p = 0.640 m), and each age group included five males and five females. All participants were free of self-reported neural or musculoskeletal disorders that would affect balance or walking. The project was approved by the Virginia Tech Institutional Review Board, and written informed consent was obtained from each participant prior to participation.

Testing involved participants walking along an 8 meter level, dry walkway covered in a low-height loop-style carpet under five gait conditions. The gait conditions included self-selected gait and four controlled gait conditions. The self-selected gait condition involved participants walking along the walkway with no instruction with respect to speed or step length. The first two controlled gait conditions controlled speed, but not step length, and involved participants walking at either 1.1 m/s (Slow) or 1.5 m/s (Fast). These speeds were chosen as representative of the range of speeds used by both younger and healthy older adults in self-selected gait reported in the literature.[12-17] Speed control was achieved by having participants match speed with a moving belt placed alongside the walkway (Figure 1). The second two controlled gait conditions controlled both speed and step length (Slow-Constrained and Fast-Constrained). The two speeds were the same as the Slow and Fast conditions, and the controlled step length was 0.65 m at both speeds. This step length represents a mid-range value for step lengths chosen by young and older adults during self-selected gait reported in the literature.[13-18] Step length was controlled by having participants step on markings on the walkway (Figure 1). All participants wore their own but similar soft-soled, closed-toe walking shoes.

The self-selected condition was performed first, followed by the four controlled gait conditions presented to each participant in a random order. Participants were allowed practice trials to ensure they were comfortable with the task in each case. For trials without controlled step length, the starting position of the participant was adjusted iteratively during practice trials so they would naturally step on the force platform without altering their chosen gait. Three trials of each controlled gait condition were recorded to increase the likelihood that a trial closely matching the target condition(s) would be recorded. Participants stepped on a six degree-of-freedom force platform (Advanced Mechanical Technology Inc., Watertown, MA) with their right foot during each trial, and ground reaction forces were sampled at 1000 Hz. Force platform data were low pass filtered at 20 Hz (4^{th} order zero-phase-lag Butterworth filter) prior to further analysis. The motions of reflective markers placed on the left and right heel and right anterior superior iliac spine were sampled at 100 Hz by a VICON 460 motion analysis system (VICON Motion Systems Inc., Lake Forest, CA).

For each trial, speed, step length, and peak required COF were determined. Speed was determined as the average forward speed of the right anterior superior iliac spine marker, and step length as the average forward distance between the heel markers during double stance phase. Required COF was determined by dividing the total shear GRF (resultant of anterior-posterior and medial-lateral force components) by the vertical GRF throughout stance phase, and identifying the peak in this ratio at about 10-20% stance time[19] (Figure 2) when the foot is supporting the majority of body weight, and when the foot would tend to slip forward. A forward slip of the foot at this point of the gait cycle is thought to be particularly dangerous[20] because it can be difficult to recover from, and thus lead to a fall. Large values of required COF that occurred at the beginning and end of stance phase due to small values of vertical GRF were considered spurious and ignored.

Required COF was analyzed using two analyses. For self-selected gait, independent *t*-tests were used to investigate differences between age groups. For the controlled gait conditions, planned contrasts after a two-way mixed-model analysis of variance were used to investigate differences between age groups when controlling speed and step length, and to investigate the independent effects of speed and step length on required COF. This two-way analysis of variance had independent variables of age group (young or older) and gait condition (Slow, Slow-Constrained, Fast, Fast-Constrained). Effects of age and gait condition on speed and step length were examined using the same analysis. The first hypothesis would be accepted if required COF differed between age groups during self-selected gait (analyzed using the independent *t*-test) and *not* differ between age groups when controlling both speed and step length (analyzed using planned contrasts between age groups for Slow-Constrained and Fast-Constrained conditions). The second hypothesis would be accepted if required COF increased between Slow-Constrained and Fast-Constrained conditions (analyzed using planned contrasts within each age group). The third hypothesis would be accepted if required COF increased between Fast and Fast-Constrained (analyzed using planned contrasts within each age group).

Results

Required COF ranged from 0.124 to 0.279 for all participants and gait conditions, with an overall mean of 0.193±0.035 (Figure 3). During self-selected gait, required COF was 13.7% lower among older adults than young (p = .031), speed did not differ between age groups (p = .162), and step length was 7.5% shorter among older adults than young (p = .019). When speed was controlled but step length was not (Slow and Fast conditions), required COF was 13.8% lower among older adults across both gait conditions (Slow: 15.4% difference and p = .030; Fast: 12.4% difference and p = .053), and step length was 5.9% shorter among older adults across both gait conditions (Slow: 6.3% difference and p = .003; Fast: 5.6% difference and p = .002). When both speed and step length were controlled (Slow-Constrained and Fast-Constrained conditions), required COF did not differ between age groups (Slow-Constrained: p = .357; Fast-Constrained: p = .941).

To investigate the independent effects of speed on required COF, required COF was compared between Slow-Constrained and Fast-Constrained gait conditions within each age group. These gait conditions differed in speed (Slow-Constrained = 1.185 m/s across both groups; Fast-Constrained = 1.526 m/s across both groups; $p < .001$), but not in step length (Slow-Constrained = 0.650 m across both groups; Fast-Constrained = 0.654 m across both groups; $p = .635$). Young adults exhibited no effect of speed on required COF when maintaining constant step length ($p = .436$), and older adults exhibited an 8.8% lower required COF during Fast-Constrained compared to Slow-Constrained ($p = .014$). To investigate the independent effects of step length on required COF, required COF was compared between Fast and Fast-Constrained within each age group. These gait conditions differed in step length (Fast = 0.763 m across both groups; Fast-Constrained = 0.654 m across both groups; $p < .001$), but not in speed (Fast = 1.523 m/s across both groups; Fast-Constrained = 1.526 m/s across both groups; $p = .772$). Young adults exhibited 33.0% higher required COF when walking with longer steps during Fast compared to Fast-Constrained ($p < .001$), and older adults exhibited 15.8% higher required COF when walking with longer steps during Fast compared to Fast-Constrained ($p < .001$).

Discussion

The first goal of this study was to investigate whether age differences in required COF during self-selected gait persisted when controlling speed and step length. Results from previous research with respect to age differences in required COF are ambiguous due to inconsistent findings, and potentially confounding differences in gait spatio-temporal characteristics between age groups.[6-9] It was hypothesized that age differences in required COF would not persist when controlling both speed and step length. Our results showed, consistent with prior studies, age differences in required COF during self-selected gait. These differences persisted when controlling speed, but were not found when controlling both speed and step length. As such, we accepted our hypothesis. These results confirm that investigations of age-related differences in required COF can be confounded by speed and step length, and that it is important to account for these gait characteristics when trying to understand the underlying factors contributing to any age-related differences in required COF (or lack thereof). Based upon these results, older adults appear to have a lower likelihood of slipping while walking compared to young adults, and this lower likelihood is due to age-related alterations in speed and step length. Our results also suggest that the increased rate of falls among older adults is not due to a greater likelihood of slipping while walking.

The second goal of this study was to determine the independent effects of speed and step length on required COF. The inter-dependence of speed and step length makes it difficult to separate and understand their independent effects. We hypothesized that increasing speed while holding step length constant would increase required COF. Our results showed that increasing speed while holding step length constant *decreased* required COF among older adults, and had no effect on required COF among young adults. As such, we rejected our hypothesis. We also hypothesized that increasing step length while holding speed constant would increase required COF. Our results showed that increasing step length while holding speed constant did indeed increase required COF. As such, we accepted our hypothesis.

The range of required COF values found here were similar to those reported in the literature.[4,5,7-10,20-22] Older adults exhibited a lower required COF compared to young adults during self-selected gait, which was similar to a previous study,[6] but differed from other studies that reported no differences in required COF between healthy older and young adults during self-selected gait.[7-9] Older adults also exhibited a lower required COF compared to young adults when gait speed was controlled, again differing from a previous study[4] that reported no differences between older and young adults during controlled slow, medium, and fast speeds without controlling step length. There are numerous possible reasons for these different findings between studies. In addition to self-selected gait speed and step lengths, other factors that differ between studies and that could influence the identification of age-related differences in required COF include footwear, experience or awareness of slipping,[22,23] and the experimental setup.

This study supports previous work indicating that older adults are not at increased risk of slipping,[4,7-9] as they did not exhibit a higher required COF compared to young adults.[4,7-94,7-9] In fact, a comparison of required COF between young and older adults without accounting for gait spatio-temporal characteristics indicated older adults had a lower required COF, which suggests a lower risk for slipping. Lockhart et al.[9] suggest that older adults are not at increased risk of slipping because they adopt a stable gait pattern with reduced speed and step length. Older adults do tend to adopt gait patterns with slower speeds and shorter step lengths than young adults,[12-14,16,18,24] and these adaptations have been associated with less severe slips when exposed to a slippery surface.[25] It has been suggested that these age differences may represent adaptations to provide a safer, more stable, gait pattern.[18] However, the current study suggests that older adults are not at increased risk of slipping even when walking with the same speed and step length as young adults.

Our results indicate speed and step length have independent and opposite effects on required COF. As speed was increased (while keeping step length constant), required COF tended to decrease among young adults, and decreased significantly among older adults. This can be seen by comparing the Slow-Constrained versus the Fast-Constrained gait conditions (Figure 3). On the other hand, as step length was increased (while keeping speed constant), required COF increased. This can be seen by comparing the Fast-Constrained versus the Fast gait conditions (Figure 3). The opposite effects of speed and step length on required COF are due to the differences in how strongly speed and step length affect shear and vertical GRFs at the same instant as the required COF, which are the numerator and denominator of required COF, respectively. Increasing speed and step length increased both shear and vertical GRFs. This is in agreement with previously reported relations between gait characteristics and GRFs.[26] Speed, however, had a larger relative effect on vertical GRF (denominator of required COF calculation) than shear GRF (numerator of required COF calculation) such that increasing speed *decreased* the required COF. For example, increasing speed from the Slow-Constrained condition to the Fast-Constrained condition (while keeping step length constant) resulted in a 29.1% increase in speed when averaged across young and older participants. This increase in speed increased shear GRF by 6.8%, increased vertical GRF by 13%, and decreased required COF by 6.1%. On the other hand, step length had a larger relative effect on shear GRF than vertical GRF such that increasing step length *increased* the required COF. For example, increasing step length from the Fast-Constrained condition to the Fast condition (while keeping speed constant) resulted in a 16.7% increase in step length when averaged across young and older participants. This increase in step length increased shear GRF by 34.3%, increased vertical GRF by 9.2%, and increased required COF by 24.4%. This quantitative example illustrates that step length has a stronger effect on required COF than speed. Speed and step length tend to be positively correlated if not controlled.[11] Increasing speed from the Slow condition to the Fast condition resulted in both a 29.1% increase in speed and a 17.3% increase in step length. Despite increasing a smaller percentage than speed, step length still had a larger effect on required COF, as illustrated by the

14.3% increase in required COF rather than a decrease that might be expected if speed had a larger effect.

Several limitations of this study warrant mention. This study was limited to healthy, community-dwelling adults walking on a level surface in their own normal walking shoes, and the results may not generalize to other conditions or populations. It has been shown that sole hardness can affect peak required COF.[27] However, all participants in the current study wore similar soft-soled, closed-toe walking shoes, and we have no reason to believe that footwear systematically affected the results. This study also used real-time visual feedback in controlling gait speed, which could have had unintended effects on gait. However, because the data suggest speed and step length were well-controlled as intended, and because participants appeared to perform the task with little difficulty, it does not seem likely that the method of gait control had a significant impact on the results of this study.

In conclusion, age differences in required COF existed during self-selected gait, but these differences did not persist when experimentally-controlling speed and step length. These results support the need to account for these gait characteristics when trying to understand the underlying factors contributing to any age-related differences in required COF (or lack thereof). Speed and step length exhibited independent and opposite effects on required COF, with step length having a strong positive effect and speed a weaker negative effect. As such, the fact that older adults typically walk with both shorter step lengths and slower speeds has the net effect of decreasing the required COF. A practical implication of these results is that the risk of slipping increases with larger steps rather than increased speed, and a faster gait with short, quick steps would not increase required COF and the risk of slipping.

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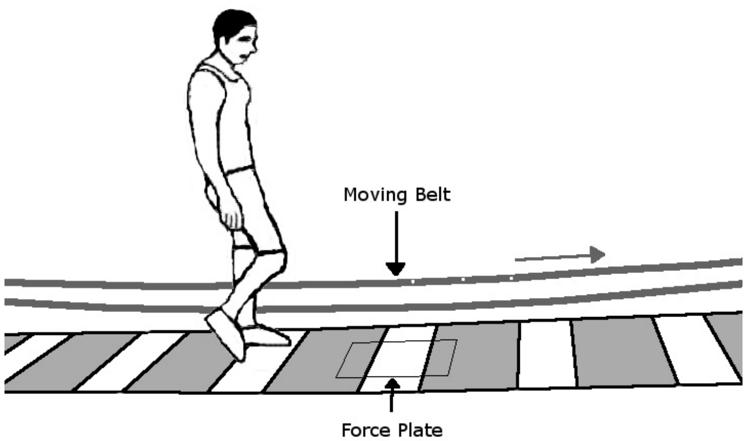

Figure 1: Walkway setup for controlled gait trials. Speed was controlled by having participants match speed with a moving belt alongside the walkway. Step length was controlled by instructing the participants to step only on the white stripes across the walkway. In the cases where step length was not controlled, the stripes were removed from the walkway.

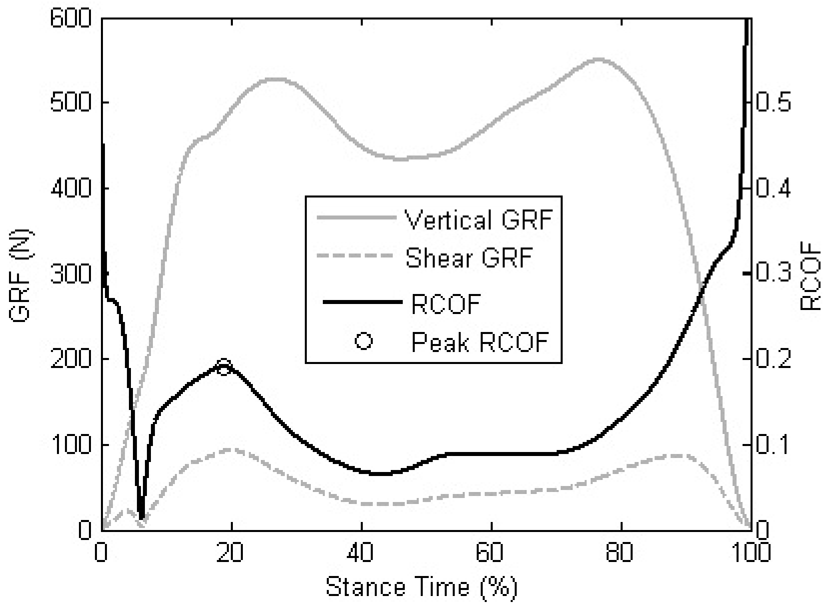

Figure 2: Required COF was calculated throughout the stance phase as the ratio of the shear GRF to the vertical GRF. Peak required COF occurred at 10-20% stance phase, when the foot would tend to slip forward (i.e. when the resultant shear GRF opposed a forward slip). Note that the shear GRF shown here is the resultant of the anterior-posterior and medial-lateral components, and thus always positive.

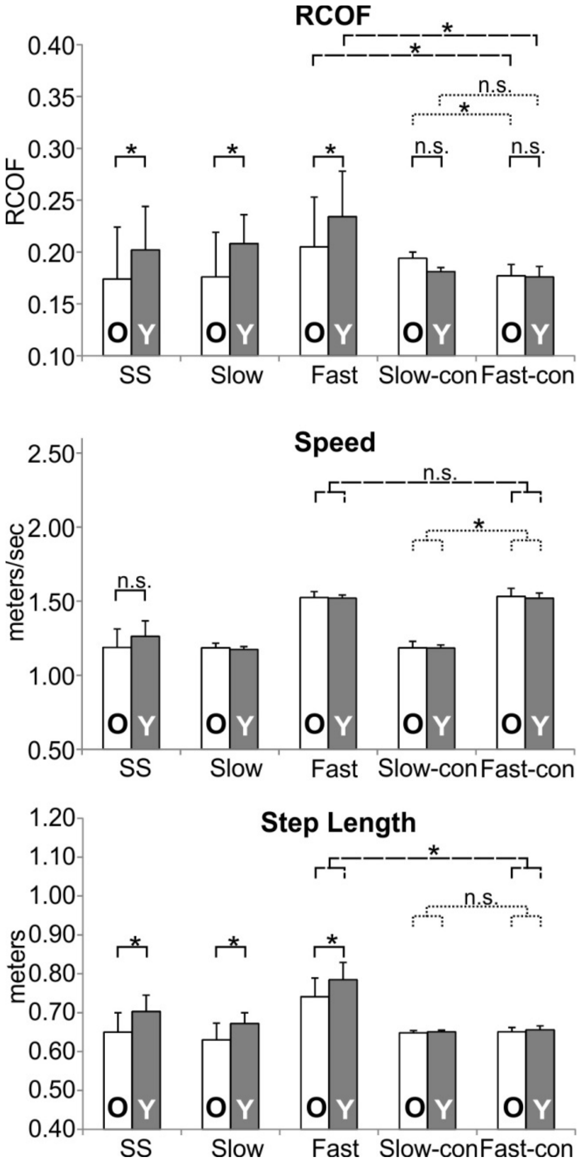

Figure 3: Mean values of required COF, speed, and step length by age group for the five gait conditions tested. Solid brackets compared between age groups within each gait condition. Dotted brackets compared between Slow-Constrained and Fast-Constrained conditions to investigate the independent effect of speed on required COF. Dashed brackets compared between Fast and Fast-Constrained conditions to investigate the independent effect of step length on required COF. * = statistically significant ($p \leq 0.05$). n.s. = not statistically significant. O = Older group. Y = young group.

Temporal changes in the required shoe-floor friction when walking following an induced slip

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Abstract

Biomechanical aspects of slips and falls have been widely studied to facilitate fall prevention strategies. Prior studies have shown changes in gait after an induced slipping event. As such, most researchers only slip participants one time to avoid such changes that would otherwise reduce the external validity of experimental results. The ability to slip participants more than once, after allowing gait to return to a natural baseline, would improve the experimental efficiency of such studies. Therefore, the goal of this study was to characterize the temporal changes in required shoe-floor friction when walking following an induced slip. Two experiments were completed, and each employed a different potential strategy to promote the return of gait to a natural baseline after slipping. In the first experiment, extended time away from the laboratory was used to promote the return of gait to baseline. We measured required coefficient-of-friction among 36 young adult male participants over four sessions. The first three sessions provided measurements during baseline (i.e., natural gait) both prior to slipping and immediately after slipping. The fourth session provided a measurement 1-12 weeks after slipping. In the second experiment, an extensive number of walking trials was used to promote the return of gait to baseline. We measured required coefficient-of-friction among 10 young adult male participants in a single session. Measurements were collected during 10 baseline walking trials, immediately after slipping, and during 50-55 additional trials. In both experiments, required coefficient-of-friction decreased 12-16% immediately after a single slip, increased toward baseline levels over subsequent weeks/walking trials, but remained statistically different from baseline at the end of the experiments. Based on these results, experiments involving slipping participants multiple times may not have a high level of external validity, and researchers are encouraged to continue to limit experimental protocols to a single induced slip per participant.

Introduction

Falls are a significant source of unintentional injuries and medical costs in the United States. In 2005, more than 8.7 million emergency department visits were made to U.S. hospitals due to fall-related injuries, making up 20% of all injury visits [1]. Additionally, the National Safety Council reported that falls are the second-leading cause of unintentional death in homes and communities, resulting in more than 25,000 fatalities in 2009 [2]. Claims for fall-related occupational injuries constitute about 25% of all workers' compensation costs in the U.S., which is estimated to total more than $6 billion annually [3]. The large number of fall-related injuries and high associated medical costs highlight the importance of research into the mechanisms and prevention of falls [4].

Slipping is responsible for a large proportion of falls [3,5-7]. For example, Courtney et al. (2001) reported that slipping contributed to 40-50% of reported fall-related injuries [3]. The causes of slips are complex, involving the interaction of numerous factors, both intrinsic (human factors) and extrinsic (environmental factors) [7-10]. The frictional properties of the interface between the shoe and floor are the primary environmental determinants of a slipping event [8,10]. In particular, the available coefficient of friction between the shoe and floor is determined largely by shoe and floor materials and environmental conditions. The required coefficient of friction (RCOF) is the minimum coefficient of friction necessary at the shoe-floor interface to prevent slipping [11], and is determined from the ratio of shear to normal ground reaction forces during stance. The local maximum of this ratio at 10-20% of stance is typically used as the RCOF, because it is at this point that slipping is thought to most likely to lead to a fall [11]. The risk of slipping increases when the RCOF approaches or exceeds the available COF [7,8,11]. However, most researchers only determine the RCOF when assessing the risk of slipping, since obtaining reliable measurements of the available COF between the shoe and floor is challenging [11]. Increases in RCOF indicate an increased risk of slipping.

The biomechanics of slips are commonly studied in laboratory settings in an effort to improve the understanding of slip mechanisms, slip and fall prevention strategies, and risk assessment methods [5,10]. For example, the RCOF has been correlated with several kinematic gait parameters [5,11-17]. One challenge in studying slips is to maintain natural gait patterns during testing in the laboratory, yet multiple studies have demonstrated that gait changes after slipping [5,10,12]. Such changes suggest that any experimental results obtained after gait has been altered due to a prior slip may have limited external validity with respect to natural gait. Therefore, most researchers only slip participants once when it is important to maintain external validity with natural walking [5,10,12,18-20]. Describing the temporal changes in gait after an induced slip may allow participants to be slipped more than once, but after an appropriate delay to allow gait to return to a natural baseline. This could substantially improve the efficiency of such experiments, as fewer participants would be needed, thereby reducing the time and resources for participant recruitment, medical screenings, and experimentation. It would also allow the use of within-subject experimental designs (or at least repeated measures for a subset of factors), which have improved statistical power over between-subject designs [21].

Therefore, the goal of this study was to characterize the temporal changes in RCOF when walking following an induced slip. RCOF was used to quantify changes in gait due to its association with risk of slipping [7,8,11]. Results from this study could aid in the experimental design of future studies involving laboratory slips, and could allow researchers to slip participants more than once, while ensuring results are descriptive of natural, unexpected slips.

Materials and Methods

Two separate experiments were completed, with a separate sample in each. All participants were young male adults recruited from the university population, and were free from self-reported musculoskeletal and neurological disorders that may affect gait or balance.

Ethics Statement

Prior to any data collection, all participants provided written informed consent by reviewing and signing a consent form that described the aims and procedures of the study. The study procedures, including the consent form, were approved by the Virginia Tech Institutional Review Board.

Experiment 1

The goal of Experiment 1 was to characterize the temporal changes in RCOF over several weeks after an induced slip. Thirty-six young adult males (mean age = 20.7 ± 2.3 years; height = 1.80 ± 0.08 m; mass = 79.8 ± 11.8 kg) were recruited from the local university population. Participants first completed three baseline sessions on separate days to determine baseline RCOF during normal walking. During the third session, and after RCOF measurements, participants were exposed to an unexpected slip. Additional walking trials were then repeated for approximately 15 minutes to assess changes in RCOF immediately after slipping. Participants then returned for a fourth session either one, two, four, six, or twelve weeks after slipping to measure RCOF. Participants were recruited in groups of three to five young adult males, and all members of each of these groups returned for the fourth session after the same number of weeks. The first group of participants returned for the fourth session one week after slipping, and subsequent groups returned after progressively more weeks. Participants in the group that returned four weeks after slipping were statistically significantly older compared to the other groups (likely because the oldest participant, aged 29, was in the group), but this group exhibited no other differences in height, mass, or baseline RCOF compared to the other groups.

At the start of the first session, participants were made aware of the possibility of an induced slip during any walking trial or session throughout the experiment. Participants donned laboratory-provided, soft-soled walking shoes to prevent variation in the frictional properties of the shoe-floor interface between participants, and wore a safety harness attached to a track above the walkway to prevent a fall. To start the experiment, participants were asked to walk at a purposeful speed (slightly faster than comfortable) along a 9 meter walkway covered in vinyl flooring. We chose this purposeful speed, rather than a slower comfortable speed, based upon our observations during pilot work that some participants walked at a slower than normal speed once they were informed that a slip may occur. Walking more slowly decreases RCOF [22] and can make it less likely that participants slip when exposed to a slippery floor. Participants were required to maintain a gait speed between 1.5 and 2 m/s during all walking trials, and trials not within a fixed speed range (-0.0525 m/s to +0.0975 m/s from the participant's mean speed) were repeated, with verbal feedback from the investigators to increase or decrease speed. Gait speed was experimentally controlled to avoid changes in speed after slipping from confounding the measurements of RCOF. Participants were given three practice trials at the beginning of each session to adjust to the environment and re-establish their gait speed from the first session (if necessary). After the self-selected purposeful gait speed was determined during the first session, data from approximately 10 acceptable walking trials (appropriate speed and foot placement with respect to a force platform) were collected. During these and all other walking trials, participants were attempting to retain a memorized set of letters, numbers, or symbols presented on note cards before each trial, to divert their attention from walking and a potential slip. Once reaching the far end of the walkway, participants were instructed to sit on a stool with their back to the walkway and memorize a new set of information until notified to turn around (approximately 1.5 minutes) and prepare for the next trial. Session two and the beginning of session three each involved data collection during approximately 10 more acceptable walking trials at the appropriate speed.

After the initial gait trials during session three, a thin layer of vegetable oil was applied with a paint roller to a middle portion of the walkway while the participants had their back to the walkway and were distracted with the memorization task. To minimize auditory or visual cues of the contaminant, participants wore noise protection earmuffs, nature sounds were played, and the lighting was dimmed throughout all sessions. Slips of the stance foot of at least 3 cm during early stance were characterized as a successful slip. If participants were unsuccessfully slipped, the walkway and shoes were cleaned and dried, and another slip was attempted after a few additional walking trials. After a successful slip trial, the walkway and shoes were cleaned and dried, restoring their original state, and 10 additional walking trials were performed. All participants who failed to slip during the first attempt were successfully slipped several trials later in a second slip attempt. A successful slip on the first attempt, or second attempt after a failed first attempt, were assumed to have the same effect on gait because in each case the participant experienced a successful slip. The fourth session was one, two, four, six, or twelve weeks after slipping, and involved approximately 10 additional walking trials.

Experiment 2

The goal of Experiment 2 was to characterize the temporal changes in RCOF over a number of walking trials performed immediately after slipping during the same session as slipping. Ten young adult males (mean age = 21.8 ± 1.8 years; height = 1.81 ± 0.07 m; mass = 76.1 ± 7.4 kg) completed this experiment. Unlike Experiment 1, these participants completed only one experimental session. Approximately ten trials were performed to determine baseline RCOF during normal walking. Participants were then exposed to an unexpected slip, followed by 50-55 additional walking trials after removing the contaminant from the floor/shoes.

At the start of the session, participants were made aware of a possible slip during any walking trial throughout the experiment. Participants donned laboratory-provided, soft-soled walking shoes to prevent variation in the frictional properties of the shoe-floor interface between participants, and wore a safety harness attached to a track above the walkway to prevent a fall. As in Experiment 1, the experiment started by asking participants to walk at a purposeful speed (slightly faster than comfortable) along a 9 meter walkway covered in vinyl flooring. The same methods were used here as in Experiment 1 to achieve and maintain a gait speed between 1.5 and 2 m/s during all walking trials. After the self-selected purposeful gait speed was determined, data from approximately 10 acceptable walking trials (appropriate speed and foot placement with respect to a force platform) were collected. As in Experiment 1, participants attempted to retain a memorized set of letters, numbers, or symbols during all trials to divert their attention from walking and a potential slip, and earmuffs were worn with background sounds and dimmed lighting. After 10 acceptable walking trials, participants were slipped as described as in Experiment 1. The walkway and shoes were then cleaned and dried, and 50-55 additional walking trials were performed. The number of post-slip trials was selected so that the entire session did not last longer than two hours.

Analysis

During each trial, the three-dimensional positions of selected anatomical landmarks were sampled at 100 Hz using a six-camera Vicon motion analysis system (Vicon Motion Systems Inc., Centennial, CO), and ground reaction forces were sampled at 1000 Hz using a force platform (Bertec Corporation, Columbus, OH). Markers were placed over the inferior tip of the right scapula, the heel and tip of each shoe, and the lateral malleolus and lateral femoral epicondyle of each lower extremity. Marker position and force platform data were low-pass filtered at 5 and 7 Hz, respectively, using an eighth-order zero-phase-shift Butterworth filter [16]. The RCOF was the primary dependent variable because it is believed to best reflect aspects of gait that contribute to the potential for slipping [5]. It was calculated from the filtered force platform data as a local maximum of the ratio of shear vs. normal ground reaction forces observed during 10-20% of the stance phase of gait [11]. Large values of RCOF that occurred at the beginning and end of stance phase, due to small values of vertical GRF, were considered spurious and ignored. Gait speed and step length were also determined. Gait speed was determined as the mean forward speed of the marker on the right scapula, and step length was determined as the distance between heel contact and contralateral-limb heel contact.

For Experiment 1, a three-way mixed-model ANOVA (independent variables were trial and session as fixed effects, and subject as a random effect) with planned contrasts was used to investigate differences in the dependent variables between sessions. Planned contrasts compared the dependent variables between the three baseline sessions (all three considered together) and each post-slip session. For Experiment 2, a two-way repeated-measures ANOVA (independent variables were trial as a fixed effect, and subject as a random effect) with planned contrasts was used to investigate differences in the dependent variables between trials. Planned contrasts compared the dependent variables between the 10 baseline trials (all 10 considered together) and groupings of five consecutive post-slip trials (first five trials after slipping, second five trials after slipping, etc.). Statistical analyses were performed using JMP 9 (SAS Institute Inc., Cary, NC) with a significance level of $p = 0.05$, and summary values are reported as least squares means ± standard error.

Results

Experiment 1

Prior to slipping, the mean RCOF across the three baseline sessions was 0.202 ± 0.003 (Figure 1). Immediately after slipping, RCOF decreased 12% to 0.178 ± 0.003 ($p < 0.001$), and exhibited a general increasing trend back toward baseline over the subsequent 12 weeks. However, all post-slip RCOF values remained statistically different from baseline ($p < 0.001$). To better illustrate the varying trends in RCOF between participants over all sessions, data from each individual participant are shown in Figure 2. All but one of the 36 participants demonstrated a decrease in RCOF immediately after slipping, and the percentage of participants who showed an increase in RCOF toward baseline increased as the number of weeks between slipping and the fourth session increased. One week after slipping, 44% (four out of nine) of participants showed an increase in RCOF toward their baseline value. Two weeks after slipping, 50% (two out of four) of participants showed an increase in RCOF toward their baseline value. Four weeks after slipping, 57% (four out of seven) of participants showed an increase in RCOF toward their baseline value. Six weeks after slipping, 73% (eight out of eleven) of participants showed an increase in RCOF toward their baseline value. Twelve weeks after slipping, 100% (five out of five) participants showed an increase in RCOF toward their baseline value.

Walking speed and step length did not exhibit any systematic trends over all sessions. Walking speed was 1.56±0.004 m/s during baseline sessions, was 0.4% faster than baseline immediately after slipping (p = 0.014), not different (p = 0.259) from baseline one week after slipping, 1.6% slower than baseline two weeks after slipping (p < 0.001), and not different from baseline for the remaining sessions (p = 0.153 at four weeks, p = 0.544 at six weeks, and p = 0.109 at 12 weeks). Step length was 0.82±0.003 m during baseline sessions, was not different from baseline immediately after slipping (p = 0.056) or one week after slipping (p = 0.224), 2.2% shorter than baseline two weeks after slipping (p < 0.001), and not different from baseline for the remaining sessions (p = 0.397 at four weeks, p = 0.112 at six weeks, and p = 0.622 at twelve weeks).

Experiment 2

Prior to slipping, the mean RCOF across the 10 baseline trials was 0.196 ± 0.007 (Figure 3). Over the first five trials after slipping, RCOF decreased 16% to 0.166 ± 0.008 (p < 0.001), and exhibited a general increasing trend back toward baseline over the 55 trials after slipping. However, all post-slip RCOF values remained statistically different from baseline (p < 0.001). To better illustrate the varying trends in RCOF between participants over all trials, RCOF for four representative participants is shown in Figure 4. Nine of ten participants demonstrated a decrease in RCOF immediately after slipping, and a linear regression fit to each participant's 50-55 trials after slipping showed a positive slope for eight of 10 participants. The two participants whose RCOF did not increase back toward baseline are shown in Figure 4. The predicted RCOF value from the eight linear regression equations with a positive slope after 55 post-slip trials averaged 95.5% of respective baseline RCOF value.

Walking speed and step length did not exhibit the same general trend as RCOF. Walking speed was 1.51 ± 0.035 m/s during baseline sessions, was up to 4.0% faster than baseline (up to 1.57 m/s) over the first 25 trials after slipping (p = 0.001-0.033), and not different from baseline over the remaining 25-30 trials (p = 0.192-0.949). Step length was 0.81 ± 0.02 m during baseline sessions, did not differ from baseline during the first five trials after slipping, and sporadically exhibited differences from baseline over the 50-55 trials after slipping (p = 0.278-0.008) with no systematic changes toward or away from baseline.

Discussion

The goal of this study was to characterize the temporal changes in RCOF when walking following an induced slip. In Experiment 1, RCOF decreased by a mean of 12% immediately after slipping, and gradually increased toward a pre-slip baseline over the next 12 weeks. However, the RCOF remained statistically lower than baseline for all follow-up sessions up to 12 weeks after slipping. In Experiment 2, RCOF decreased by a mean of 16% immediately after slipping, and gradually increased toward baseline over the next 50-55 trials. However, the RCOF remained statistically lower than baseline for all of these trials. These results indicate that: 1) waiting 12 weeks after slipping, for a potential follow-up experimental session to retest participants, is not sufficient for gait to return to baseline; and 2) repeating 50-55 walking trials after slipping during the same experimental session is not sufficient for gait to return to baseline.

The observed 12-16% decrease in RCOF immediately after slipping is consistent with prior studies and suggests the use of a more cautious gait to reduce the risk of slipping. Cham and Redfern (2002) reported a 5-12% decrease in RCOF from baseline (mean baseline RCOF = 0.18) after slipping while walking over level ground, even though participants were assured they would not be slipped again [5]. Lockhart et al. [13] and Siegmund et al. [23] reported a 20% and 7% decrease in RCOF, respectively, after exposure to a slip. Changes in gait speed and step length immediately after slipping have not been consistently reported. Cham and Redfern (2002) reported no change in stride length after slipping, and gait speed was neither controlled nor reported [5]. Lockhart et al. (2007) reported an 18% decrease in step length, asked participants to walk at their self-selected speed during all trials, and did not report results with respect to gait speed. Heiden et al (2006) reported no change in step length or gait speed after experiencing a slip [12]. We chose to control gait speed to within a small range of each participant's self-selected speed because participants slowed their gait after slipping during pilot testing, and we did not want changes in gait speed to confound our results with respect to RCOF. Controlling speed, however, may have mitigated changes in step length compared to other studies.

Across both Experiments, RCOF decreased an average of 0.028, or 14.1%, immediately after slipping. A change in RCOF of this magnitude can have a substantial effect on the probability of slipping, depending on how close RCOF and the available COF (ACOF) are in magnitude. Burnfield and Powers (2006) demonstrated that it is possible to predict a slip event based on the difference between the ACOF and RCOF [24]. They reported a 5% probability of a slip occurring when the RCOF was 0.047 lower than the ACOF, and a 50% probability of a slip occurring when the RCOF was 0.006 greater than ACOF [24]. These authors also note that, depending upon ACOF, an increase in RCOF as small as 0.05 can contribute to a substantial increase in risk of slipping. Therefore, the mean decrease in RCOF of 0.028 observed in this study would seem to be practically relevant in terms of the risk of slipping.

Experiments 1 and 2 were designed to characterize not only the temporal changes in RCOF after slipping, but also to serve as sample experimental designs that could be employed in future studies in which participants are exposed to more than one slip while allowing for a sufficient "wash out" period for gait to return to a natural baseline. With this in mind, both experimental designs aimed to maximize experimental efficiency by minimizing the number of sessions required from each participant. Experiment 1 aimed to allow RCOF to return to baseline during time away from the lab, and in application would only require one experimental session for each slip (with the necessary time for RCOF to return to baseline between consecutive sessions). Experiment 2 aimed to allow RCOF to return to baseline through repeated trials during a single experimental session, and in application would only require one experimental session for multiple slips (with the necessary number of trials for RCOF to return to baseline between slips). An alternative experimental design we considered, but elected not to pursue, would have required participants to complete multiple post-slip sessions at shorter intervals (1-2 days apart) until RCOF returned to baseline. This design may have resulted in RCOF returning to baseline more quickly than we found in Experiment 1, but would be inefficient in that it would likely require multiple experimental sessions for RCOF to return to baseline.

The RCOF of some participants returned to near baseline, while others did not (Figures 2 and 4). Given this inter-participant variability, it may be possible to track each participant's temporal changes in RCOF, and perform repeated slips on only those whose RCOF returned to baseline. However, we did not pursue this approach because we felt it to be experimentally inefficient since the testing on many participants whose RCOF does not return to near baseline would contribute to wasted time and resources. Future studies could better evaluate the costs and benefits of this approach.

The results from Experiment 1 showed that post-slip gait adaptations indicative of a lower risk of slipping persisted for at least 12 weeks (although the magnitude of these adaptations waned with time). While this time duration can be viewed as a challenge for researchers desiring to slip participants multiple times, it provides support for training interventions for improved slip prevention. Researchers have proposed that slipping or tripping individuals periodically in a controlled environment is a training intervention that can alter gait to reduce the risk of slips and trips, or improve balance recovery ability [25-27]. Our results indicate that changes in gait induced by a single slip last up to 12 weeks, which suggests that a slip or trip training intervention may also have lasting benefits.

Several limitations of this study warrant discussion. First, it is unclear if the slip-induced changes in RCOF observed in a laboratory setting generalize to outside of the laboratory. The informed consent process used in this study made participants aware of a potential slip, which may have a heightened their awareness of slipping more than is typical in a natural environment. As such, it is possible that the changes in RCOF were limited to the laboratory. Second, this study only investigated changes in gait after slipping. Therefore, no conclusions can be made about the temporal changes in balance recovery capability following a slip. Third, this study was limited to young adult males to avoid potential age and gender effects. It is unclear if the temporal changes in RCOF would differ among other populations. Fourth, we investigated the temporal changes in RCOF after a single slip. It is unclear if the temporal changes in RCOF after an additional slip would differ from after the initial slip.

In conclusion, RCOF during gait decreased 12-16% immediately after a single slip, tended to return toward baseline over subsequent weeks/walking trials, but remained statistically different from baseline at the end of our experiments. Given these results, experiments involving slipping participants multiple times do not appear practical at this time, and slip researchers are encouraged to continue to limit slips to one per participant to maintain high external validity of their results. Although other experimental designs involving more sessions per participant may help to induce RCOF to return to baseline more quickly, these would likely be less efficient and thus take away from the benefits of slipping research participants more than once. Our results also provide support for the use of perturbation-based balance training for reducing fall risk, as the changes in RCOF, which suggest a lower risk of slipping, persisted for an extended period of time.

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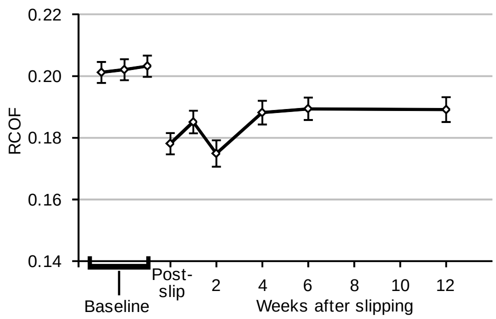

Figure 1. Required coefficient-of-friction (RCOF) across all participants of Experiment 1. Least-square means are shown, and error bars indicate standard error. RCOF was significantly different from baseline at all times after slipping. Post-slip indicates RCOF values immediately after slipping (during the third baseline session).

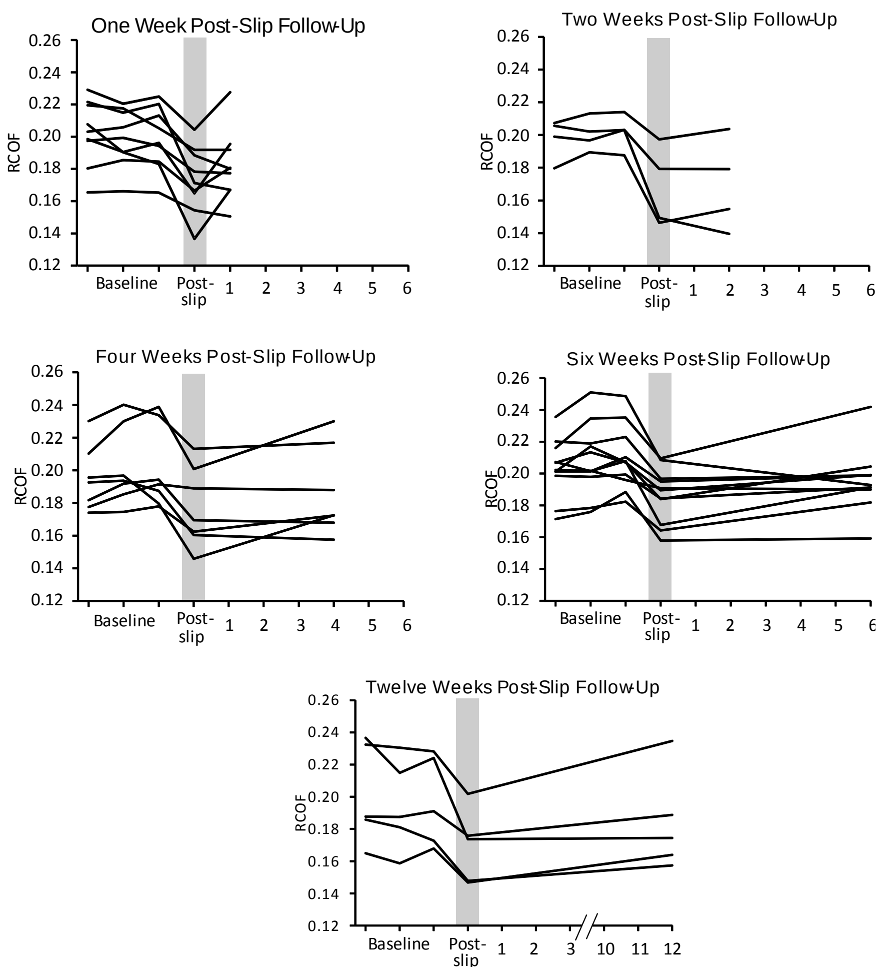

Figure 2. RCOF trends for each participant in Experiment 1. These plots illustrate the variability of RCOF values and temporal changes between participants. As the time after slipping increased, the percentage of participants who exhibited an increase back toward baseline increased. Post-slip indicates RCOF values immediately after slipping (during the third baseline session).

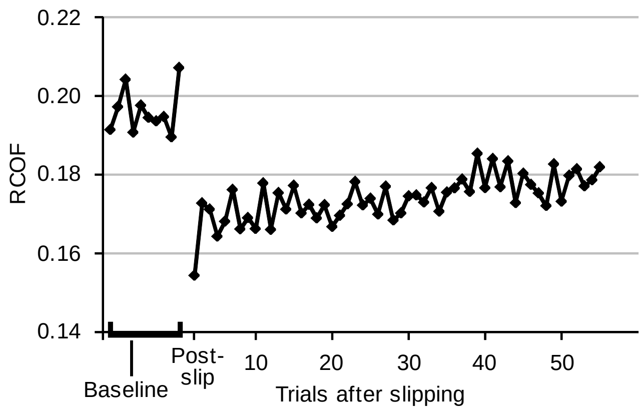

Figure 3. RCOF (least square means) across all participants in Experiment 2. Error bars not included for clarity. RCOF was significantly different from baseline for all trials after slipping. Post-slip indicates RCOF values immediately after slipping.

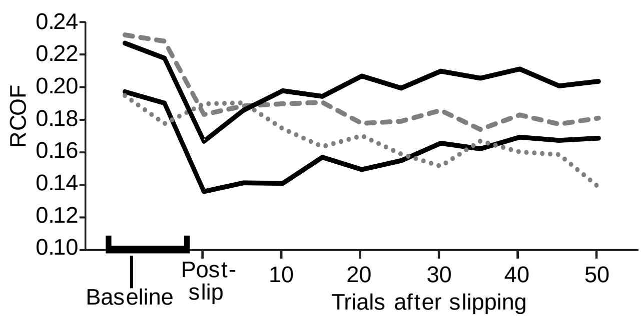

Figure 4. RCOF (least square means) for four representative participants in Experiment 2. Each data point is a mean across five consecutive trials (trials 1-5 after slipping, trials 6-10 after slipping, etc.). The two participants whose RCOF values did not trend back toward baseline after slipping are shown by dotted and dashed lines. Post-slip indicates RCOF values immediately after slipping.

A bootstrapping method to assess the influence of age, obesity, gender, and gait speed on probability of tripping as a function of obstacle height

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Abstract
Tripping is a common mechanism for inducing falls. The purpose of this study was to present a method that determines the probability of tripping over an unseen obstacle while avoiding the ambiguous situation wherein median minimum foot clearance (MFC) and MFC interquartile range concurrently increase or decrease, and determines how the probability of tripping varies with potential obstacle height. The method was used to investigate the effects of age, obesity, gender, and gait speed on the probability of tripping. MFC was measured while 80 participants walked along a 10-meter walkway at self-selected and hurried gait speeds. The method was able to characterize the probability of tripping as a function of obstacle height, and identify effects of age, obesity, gender, and gait speed. More specifically, the probability of tripping was higher among older adults, lower among obese adults, higher among females, and higher at the slower self-selected speed. Many of these results were not found, or clear, from the more common approach on characterizing likelihood of tripping based on MFC measures of central tendency and variability.

Introduction
Fall-related injuries among older adults are a major public health problem due to their high medical costs and negative impact on quality of life (Bruce et al., 1992). Tripping accounts for 35-53% of these falls (Berg et al., 1997; Blake et al., 1998). The most common measure for characterizing the probability of tripping while walking is the minimum foot clearance (MFC) during swing. A decrease in the central tendency (i.e. mean/median) of MFC, or an increase in MFC variability, are both associated with an increased probability of tripping (Begg et al., 2007; Mills et al., 2008; Winter, 1992). These indirect measures of probability of tripping, however, can lead to ambiguous results when both increase or decrease simultaneously. For example, Nagano *et al.* (2011) reported higher median MFC and MFC interquartile range (IQR, a measure of MFC variability) during overground walking compared to treadmill walking (Nagano et al., 2011), and Rossi *et al.* (2013) reported higher median MFC and MFC IQR in the non-dominant leg and at faster gait speeds (Rossi et al., 2013). Median MFC and MFC IQR are also positively correlated (Begg et al., 2007), indicating concurrent increases or decreases in both are to be expected.

The purpose of this study was to present a method that determines the probability of tripping over an unseen obstacle while avoiding the ambiguous situation wherein median MFC and MFC IQR concurrently increase or decrease, and determines how the probability of tripping varies with potential obstacle height. The method was used to investigate the effects of age, obesity, gender, and gait speed on the probability of tripping. These factors were investigated based upon reports of elevated risks of falling and fall-related injuries among adults over the age of 65 (Bruce et al., 1992; Kannus et al., 1999), individuals who are obese (Fjeldstad et al., 2008; Himes and Reynolds, 2012; Patino et al., 2010), females (Ambrose et al., 2013; Stevens, 2005), and changes in risk of tripping with gait speed (Rossi et al., 2013; Schulz, 2011).

Methods
Eighty participants completed the study including four gender-balanced groups comprised of 20 participants each (Table 1). None of the participants self-reported a change in body mass of >2.3 kg over the six months prior to testing, or any musculoskeletal, neurological, or balance disorders that would affect gait. The study was approved by the university Institutional Review Board, and all participants provided written informed consent prior to participation.

Participants walked along a 10-meter level walkway at a self-selected speed (always performed first) and a hurried speed of 1.9 m/s. Eight trials at each speed were completed, and data obtained from each trial included the swing phase of both the dominant and non-dominant leg. Thus, 16 swing phases were analyzed from each participant at each speed. The positions of three reflective markers attached to the shoe were sampled at 100 Hz using a Vicon MX motion analysis system (Vicon Motion Systems Inc., LA, CA). Multiple virtual points on the sole of the shoe were tracked using a method described elsewhere (Startzell and Cavanagh, 1999), and MFC was defined as the lowest of all points near mid-swing in a given swing phase.

MFC values were used to create trip probability curves that indicated how the probability of tripping varied as a function of height of a potential tripping obstacle (Figure 1). For potential tripping obstacle heights ranging from 0 - 7 cm, in increments of 2 mm, each experimental MFC value was dichotomized as either a trip (if the potential obstacle height was greater than MFC) or a non-trip (if the potential obstacle height was equal to or less than MFC). The percentages of trips at each obstacle height were then computed, serving as an estimate of the probability of tripping.

A statistical bootstrapping technique (Duhamel et al., 2004), was then used at each potential obstacle height to determine whether the probability of tripping differed by age group, obesity group, gender, or gait speed condition. The first step in this technique was to randomly reassign group labels to each of the 16 MFC values from each participant (e.g. young or older when investigating age effects). A probability curve was then created for each group, and the difference in trip probability between groups was calculated at each potential obstacle height. This process was performed 100,000 times to obtain a distribution of differences at each potential obstacle height that would occur if group assignment was random. This distribution acted as the sampling distribution of differences under the null hypothesis that the groups had equal trip probabilities.

The second step in this technique was to determine whether the actual observed difference in probability of tripping between groups was statistically significant. The actual observed difference in probability of tripping between groups was defined as the absolute value of the difference between the group percentages at a potential obstacle height. Because each bootstrapping analysis involved 36 comparisons between groups (0-7 cm obstacles heights in increments of 2 mm), the significance level was 0.05/36, or α=0.0014, to avoid consequences of type I errors. As such, if the actual difference in probability of tripping was in the outer 0.14% of the distribution, then the difference in trip probability between groups was considered statistically significant. Alternatively, the percentage of the distribution of differences outside of the actual observed difference yielded a bootstrap *p*-value. This second step was performed for all potential obstacle heights, and between all participant groups of interest, to determine the specific heights at which the probability of tripping differed between groups.

Group differences identified from this statistical bootstrapping technique were compared with group differences identified using the traditional measures of median MFC and MFC IQR. Group differences in median MFC and MFC IQR were determined using a four-way, mixed-factor analyses of covariance (ANCOVA) with planned contrasts. Independent variables in the ANCOVAs were age, obesity, and gender, and gait speed was the covariate. Analyses were performed using JMP v7 (Cary, North Carolina, USA).

Results

Age-related differences in the probability of tripping were not consistent between the bootstrapping technique and the ANCOVA analysis. Among normal-weight adults (Figure 1a), the probability of tripping was lower among older adults across a range of obstacle heights (2.0-4.6 cm), while no age effects were found for either median MFC or MFC IQR. Among obese adults (Figure 1b), the probability of tripping was also lower among older adults, but across a smaller range of obstacle heights (1.2-2.4 cm), while again there were no age effects for either median MFC or MFC IQR.

Obesity-related differences in the probability of tripping were also not consistent between the bootstrapping technique and the ANCOVA analysis. Among older adults (Figure 1c), the probability of tripping was significantly higher among obese adults across a range of obstacle heights (2.4-4.2 cm), while there were no obesity effects on median MFC or MFC IQR. Among young adults (Figure 1d), there were no significant effects of obesity on the probability of tripping, nor on median MFC or MFC IQR. With respect to gender (Figure 1e), the probability of tripping was higher among females across a range of obstacle heights (0.8-4.4 cm), while both median MFC and MFC IQR were lower among females. With respect to speed (Figure 1f), the probability of tripping was lower for the faster hurried speed across a narrow range of obstacle heights (4.2-5 cm), while both median MFC (approached significance) and MFC IQR were higher at the faster hurried speed.

Discussion

While prior work has employed median MFC and MFC IQR as indirect measures of likelihood of tripping, the method presented here directly determines the probability of tripping as a function of obstacle height, and uses a statistical bootstrapping technique to compare this probability between groups of interest. This technique identified effects of age and obesity that were not identified from the more traditional approach using ANOVA. This new technique also identified effects of gender and gait speed, and helped clarify ambiguous results from the ANCOVA analysis with respect to probability of tripping (e.g. when both median MFC and MFC IQR were higher among males compared to females).

Three limitations to the method presented here warrant mentioning. First, this method, along with ANOVA using median/mean MFC and MFC IQR, focuses on foot clearance at the instant that MFC occurs, even though a trip could occur at other instances during the swing phase. Second, unlike an ANOVA based upon median/mean MFC and/or MFC IQR, the current method cannot incorporate measures of covariance, or statistically control for the effects of other variables, when evaluating an independent variable of interest. Third, this method, along with most other investigations of MFC, assumes individuals will not see or react to an obstacle in their path. While this may be true for smaller obstacles, this is less likely for larger obstacles.

The method presented here may be helpful in ensuring that safety guidelines are inclusive and protective for diverse populations. For instance, The Americans with Disabilities Act (ADA) stipulates that abrupt changes in height of a walkway greater than 6 mm require edge treatment to account for individuals in wheelchairs and individuals whose foot is impeded during the swing phase of gait (Cohen and Abele, 2007). The results in Figure 1c indicate that trip probability does not differ between normal-weight and obese older adults unless obstacle height exceed 2.4 cm, suggesting that the 6 mm standard of the ADA is equally protective for both of these populations.

A statistical modeling technique reported by (Best and Begg, 2008) also characterizes the probability of tripping over a range of obstacle heights. While this modeling technique helps recognize the features of the distribution of MFC data (i.e skewness and kurtosis), the method reported here may provide a pragmatic alternative for characterizing the probability of tripping. Of note, though, is that trip probabilities obtained from the two methods differed substantially. For an obstacle height of 1 cm, Best and Begg (2008) reported a trip probability of 50% (Best and Begg, 2008) whereas the current method yielded a probability of less than 5% (depending upon the specific group of interest). This difference may be due to methodological differences between the two studies including overground vs treadmill walking, walking speeds, number of participants, and methods used to estimate MFC.

The method presented here identified differences in probability of tripping that provided both novel and complementary insight to the literature. Regarding obesity, no studies to our knowledge have reported effects of obesity on median/IQR MFC. The higher probability of tripping among obese individuals found here (albeit only among older adults) may help to explain the higher fall rates among individuals who are obese (Fjeldstad et al., 2008). Regarding gender, lower median MFC among females reported by Rossi et al. (2013) is consistent with the median MFC results presented here. However, the concurrently lower MFC IQR among females also reported here obscures the net effect of these differences on probability of tripping. The current method provides clear and direct biomechanical evidence that the probability of tripping is higher among females. Interestingly, these results appear to align with those by Berg et al., who in a one-year prospective survey study of adults aged 60-88, found that falls among females most often occurred due to tripping whereas falls among males most often occurred due to slipping (Berg et al., 1997). Regarding age, the current method identified a lower probability of tripping among older adults, but no age differences in median/IQR MFC. The majority of MFC studies reviewed by Barrett et al. (2010) also found no age differences in median MFC, but did find higher MFC IQR which suggests a higher probability of tripping among older adults (Barrett et al., 2010). Thus, the results of the current method with respect to probability of tripping are consistent with these findings. The present method also has the advantage of identifying the specific obstacle heights at which age differences exist, which may be helpful for various trip and fall prevention strategies.

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Conflict of interest statement
The authors have no conflicts of interest to report.

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Table 1: Participant demographics (mean ± standard deviation)

	Young		Older	
	NW	OB	NW	OB
Sample Size	F: (n=10) M: (n=10)	F: (n=10) M: (n=10)	F: (n=10) M: (n=10)	F: (n=10) M: (n=10)
Age (years)	F: 24.4 ± 3.4 M: 23.8 ± 3.2	F: 24.8 ± 2.8 M: 21.9 ± 2.5	F: 66.8 ± 4.9 M: 65.8 ± 4.6	F: 65.6 ± 5.5 M: 74.3 ± 6.1
BMI (kg/m^2)	F: 23.1 ± 2.2 M: 21.2 ± 1.7	F: 34.0 ± 3.5 M: 33.2 ± 3.1	F: 23.8 ± 2.0 M: 24.5 ± 1.4	F: 33.1 ± 2.0 M: 31.5 ± 1.7

Note: NW = normal-weight group, OB = obese group, F = female, M = male

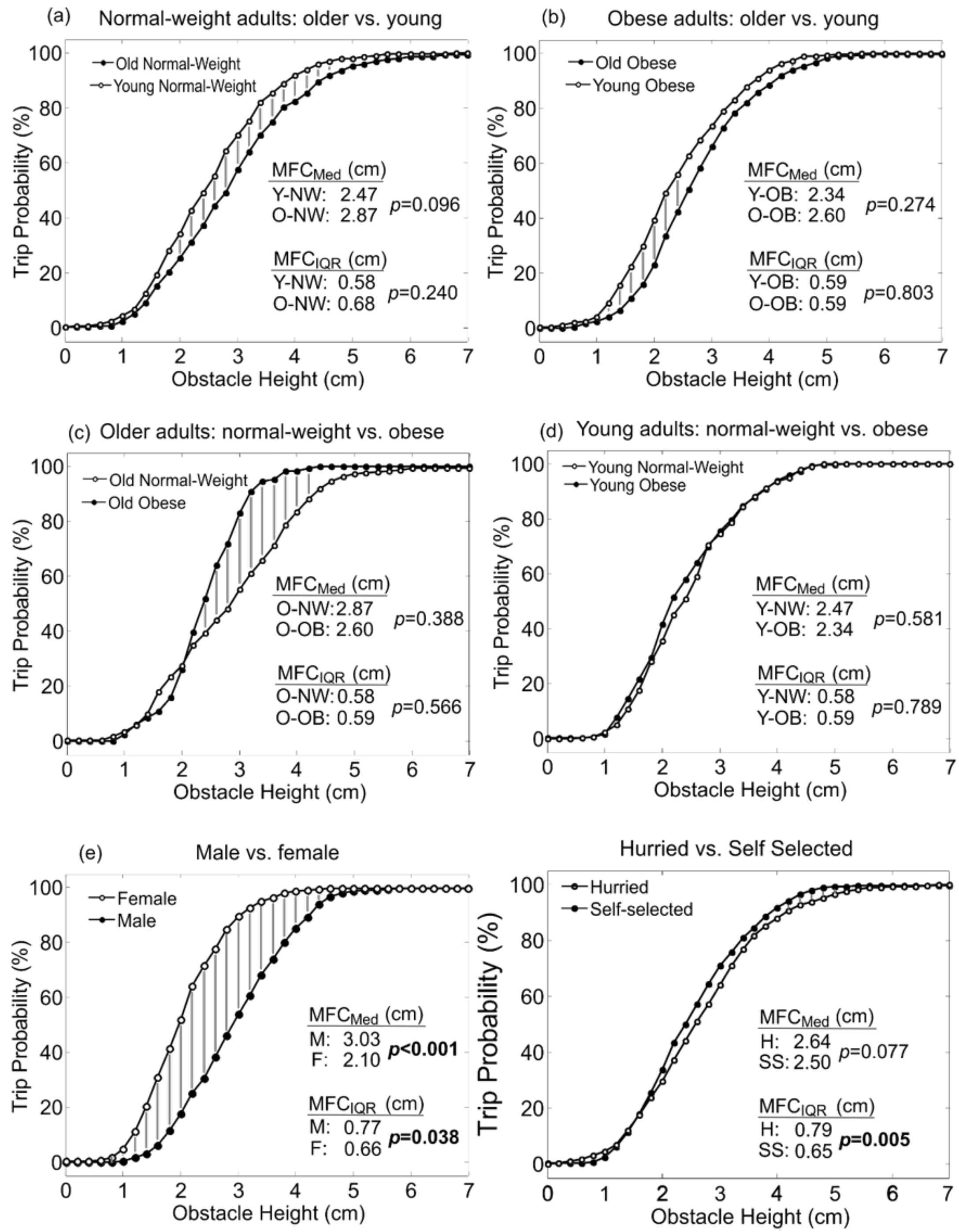

Figure 1: Trip probability curves and median/IQR MFC separated by age group, obesity group, gender, and speed. Differences in probability of tripping between groups ($p<0.0014$) are indicated by a solid vertical line, and differences in median/IQR MFC ($p<0.05$) are indicated by bold.

EXECUTIVE FUNCTION AND MEASURES OF FALL RISK AMONG PEOPLE WITH OBESITY[1]

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Summary
This study investigated the effects of obesity on executive function among young adults, and between executive function and fall risk among the obese (as estimated from select gait parameters). Thirty-nine young adults (age=21.3±2.6 years) were recruited from the local university population, including 19 being obese (based upon BMI and body fat percentage) and 20 non-obese. Four major components of executive function were assessed including selective attention, divided attention, semantic memory, and working memory. Participants performed single- and dual-task walking (walking-while-talking) to evaluate fall risk during gait as measured by minimum toe clearance, required coefficient of friction, stance time, and stance time variability. Obese participants exhibited lower scores for selective attention, semantic memory, and working memory. All participants exhibited gait changes suggestive of a higher fall risk (e.g. lower minimum toe clearance, longer stance time, and increased stance variability) during dual-task walking compared to single-task walking, and executive function scores (selective attention) were associated with gait (stance time variability) during dual-task walking. Results indicate obesity adversely affects executive function among young adults, which may be associated with their increased fall risk.

Introduction

Over the past four decades, the prevalence of obesity in the United States has more than doubled from 15% to 36% (Flegal, Carrol, Kucamarski, Johnson, 1998; Flegal, Carroll, Ogden, Curtin, 2010). This is problematic because obesity is associated with numerous maladies. Obesity is also associated with an increased risk of falling in that people who are obese are reported to fall almost twice as often (27% vs. 15%) as people who are not obese (Fjeldstad, Fjeldstad, Acree, Nickel, Gardner, 2008). Falls have also been identified as the most common (~36%) cause of injuries among people who are obese (Matter, Sinclair, Hostetler, Xiang, 2007). However, the underlying reasons for the higher fall rate among people who are obese remains unclear.

Executive function is an umbrella term for cognitive processes that regulate, control, and manage other cognitive processes (Elliott, 2003) such as attention, planning, memory, problem solving, concentration, and task-switching. Executive function appears to play an essential role in appropriately allocating attention between walking and other tasks to maintain balance and prevent falls. To this end, a dual-task paradigm involving walking and simultaneously performing another task has been used to assess attention allocation between gait/balance control and a secondary task (Aldridge, 2009; Hyong, 2015; Schaefer & Lindenberger, 2013). Schaefer and Lindenberger (2013), for example, found gait variability (i.e. single support time) increased when a working memory task was added to walking, perhaps suggesting reduced psychomotor function due to attention being divided between two tasks (Brach, Wert, VanSwearingen, Newman & Studenski, 2012). Another dual-task paradigm involving walking while talking has also been used as a test of divided attention, especially in the context of identifying fall risks. Coppin et al. (2006) reported an association between executive function and the decline in walking speed when adding talking to walking. Holtzer et al. (2014) reported a greater decline in walking speed when adding talking to walking among older adults who exhibited worse executive function. Verghese et al. (2007) reported a decrease in walking speed, cadence, and step length, and an increase in double support time when adding talking to walking, and these gait changes were attributed to dividing attention between walking and talking tasks. These studies indicate lower scores on executive function assessments are associated with larger changes in gait when adding a secondary task to walking.

Prior studies support an adverse effect of obesity on executive function. A recent review of 31 studies reported that children (up to 18 years old) who were obese performed significantly worse on tests for executive function compared to healthy weight controls (Reinert, Poe & Barkin, 2013). No studies to our knowledge have investigated a similar effect among adults. The mechanism by which obesity adversely affects executive function remains unclear. However, the cerebellum and cerebral cortex play key roles in executive function (Golfman-Rakic, 1998; Mesulam, 1998), and alterations in these structures have been associated with obesity (Miller, Couch, Schwenk, Long, Towler & Theriague, et al., 2009)(Gustafson, Lissener, Bengtsson, Bjorkelund & Skoog, 2004)(Raji, Ho, Parikshak, Beck & Lopez, 2010).

The higher fall rate among people with obesity may be due, in part, to the adverse effects of obesity on executive function and subsequent changes to gait. Obesity elicits several changes in gait including slower walking velocity (Hills & Parker, 1991; Lai, Leung, Li & Zhang, 2008; Spyropoulos, Pisciotta, Pavlou, Carins & Simon 1991; Wearing, Henning, Byrne, Steele & Hills, 2006), longer stance time (Hills & Parker, 1991; Lai, et al., 2008; Wearing, et al., 2006), and longer double stance time (Hills & Parker, 1991; Lai, et al., 2008; McGraw, McClenaghan, Williams, Dickerson & Ward, 2000;). Obesity also elicits changes in gait that are associated with an increased risk of losing balance or falling, including higher required coefficient of friction (RCOF) that suggests a higher likelihood for slipping (Wu, Lockhart & Yeoh, 2012), lower minimum toe clearance (MTC) that suggests a higher likelihood for tripping (Garman, Franck, Nussbaum & Madigan, 2015), and longer stance time that suggests compromised balance stability (DeVita & Hortobagyi 2003). Additional changes in gait would seem possible during dual-task conditions when attentional demands are higher (Forhan & Gill, 2013; Gill, 2011; Hung, Gill & Meredith, 2013), particularly since people who are obese appear to exhibit altered executive function. These additional changes in gait could further increase risk of falling.

The purpose of this study was two-fold: to investigate whether there is a negative effect of obesity on executive function among young adults, and to determine whether there is a relationship between executive function and fall risk among obese young adults. Four hypotheses were tested: 1) executive function would be adversely affected by obesity; 2) adding a dual-task to walking would increase fall risk among both obese and non-obese participants; 3) the increase in fall risk associated with adding a dual-task to walking would be more substantial among obese participants compared to non-obese participants; and 4) executive function would be inversely associated with fall risk. Executive function was assessed using standardized tests (Lezak, et al., 2004) including the Stroop test (for selective attention), Trail Making test (for divided attention, visuomotor tracking, and cognitive flexibility), Verbal Fluency test (for semantic memory) and Digit-span test (for working memory). Fall risk was assessed using measures of likelihood of slipping (RCOF), likelihood of tripping (MTC), and overall gait stability (stance time and stance time variability).

Methods

Thirty-nine young adults completed the study including 19 obese (14 females and 5 males; body mass index or BMI = 33.0 ± 2.9 kg/m^2; body fat percentage range = 5.5 - 35%; education = 15.0 ± 1.5 years) and 20 non-obese (14 females and 6 males; BMI = 22.2 ± 2.2 kg/m^2; body fat percentage range = 25.6 - 43.2%; education = 15.2 ± 1.7 years). There was no difference in age (t = 0.138, p = 0.891) or education (t = 0.503, p = 0.0.618) between obese and non-obese groups. Inclusion criteria required obese participants to have a body fat percentage above 35% for women and above 25% for men (WHO, 1995), and BMI ≥ 30 kg/m^2. Body fat percentage was estimated using skinfold calipers and measurements at three locations (on the front of the upper arm, on the back inferior to the scapula, and on the abdomen 1 cm lateral to the navel)m and all measurements were taken by the same investigator. Participants were also required to be free from any self-reported foot pain or conditions (such as previous stroke, Parkinson's disease, pinched spinal nerve and history of detached retina) that might have affected their performance in this study. This study was approved by the local institutional review board, and all participants provided written informed consent prior to testing.

Participants completed three experimental tasks during one experimental session. These tasks included 1) assessing executive function, 2) performing single-task walking (so all attentional resources could be dedicated to walking), and 3) performing dual-task walking (requiring attention to be divided). Four tests of executive function were performed including the Stroop test, Trail Making test, Verbal Fluency test, and Digit-span test (Lezak, et al., 2004). These four tests were selected due to their ability to assess different aspects of executive function, and their sensitivity to obesity effects (Reinert, et al., 2013). The Stroop test included the color-word naming subtest. Performance was assessed by the time required to name 100 items, with a shorter time indicating better performance. The Trail Making test consisted of 25 circles on a piece of paper, and the circles included both numbers (1-13) and letters (A-L). Participants were asked to draw lines to connect the circles in an ascending pattern, with alternating letters between the numbers (i.e. 1-A-2-B-3-C). Performance was assessed by the time required to complete the test, with a shorter time indicating better performance (three participants, 1 obese and 2 non-obese, made mistakes during the task, so they were excluded from all further analyses involving the Trail Making test). The Verbal Fluency test involved two subtests, with participants asked to name as many words as possible that start with a given letter (i.e., words starts with letter 'p') and in a given category domain (i.e., fruit), each for 60 seconds (Rosen, 1980). A higher number of correct words indicated better performance. The Digit Span test involved two subtests, including forward and backward tests (Foster, Lidder & Sunram, 1998). For both subtests, participants were presented with a series of digits, and they were required to immediately repeat them back. The series of digits were presented verbally to participants at a rate of one per second. For the forward subtest, the participant's task was to repeat each sequence exactly as it was given. For the backward subtest, the participant was asked to repeat each sequence in reverse order. Both subtests began with three digits and increased by one at a time up to nine. Each subtest was performed twice using different series of digits. The number of successful sequences was considered the score, and ranged from 0 to 14.

The difference between the forward and backward subtest scores was used as an indicator of working memory function, with smaller differences indicated better working memory (Foster, et al., 1998). All executive function tests were conducted by the same investigator.

Single-task and dual-task walking were then performed with the order counterbalanced across participants within each group. All participants wore a T-shirt, tight-fitting shorts, and identical dress shoes (hard PVC soles) to minimize shoe-sole differences. For single-task walking, participants were instructed to walk at a self-selected speed along a 9-m walkway. For dual-task walking, participants were instructed to walk at a self-selected speed along the walkway, but while also reciting every other letter of the alphabet (Verghese, et al., 2007). Participants were instructed to “pay equal attention to both walking and talking” and “if the first 13 letters are finished, continue with the second 13 letters, starting with a letter ‘B’ if starting with letter ‘A’ the first time and vice versa, and continue this process until instructed to stop.” The initial letter on the reciting task was randomly varied between “A” (A-C-E) and “B” (B-D-F) between participants. To reduce learning effects, participants were given three minutes to practice reciting alternate letters of alphabet, and 10 trials to practice the dual-task walking, prior to data collection. Practice walking trials also allowed participants to acclimate to the experimental setup, and allowed for determining the appropriate starting point along the walkway (so that participants naturally and consistently stepped on a force platform embedded within the walkway). Five walking trials were performed during each task.

During walking trials, ground reaction forces were sampled at 1000 Hz using a six degree-of-freedom force platform (Advanced Mechanical Technology Inc., Watertown, MA, USA) in the center of the walkway. The position of reflective markers on the left and right calcanei, top of the feet, and one on the right upper back were sampled at 100 Hz using a six-camera motion analysis system (MX-T10, VICON Motion Systems Inc., Lake Forest, CA, USA). Both the forceplate and markers data were low-pass filtered (4th order Butterworth, zero-lag filter), using 40 and 6 Hz cutoffs, respectively.

Several dependent variables were calculated from data in each walking trial, and were selected because they are related to balance control and fall risk among the elderly and people with obesity (Wu, Lockhart & Yeoh, 2012; Yogev-Seligmann, et al., 2008). MTC is associated with the likelihood of tripping while walking (Garman, et al., 2015), and was calculated as the minimum vertical distance between the toe marker and the ground during mid-swing phase of the gait cycle (Khandoker, et al., 2010). The RCOF is associated with the likelihood of slipping while walking (Wu, et al., 2012), and was calculated as the maximum ratio of resultant shear ground reaction force to the vertical ground reaction force (Wu, et al., 2012). Heel contact and toe off were identified from heel and toe kinematics (Meckelborough, van der Linden, Richards & Ennos, 2000), and stance time was the period between these two events for a single limb (Taylor, Delbaere, Mikolaizak, Lord & Close, 2013). The interquartile range (IQR) of stance time was used to quantify the variability of stance, and was computed for each participant over 10 steps.

To address hypothesis one, a one-way analyses of covariance (ANCOVA) was performed on Stroop Test time, Trail Making Test time, the number of words named in the Verbal Fluency tests (both letter and category conditions), and Digit Span scores. The independent variable for this analysis was group (obese or non-obese), and covariates included age and years of education. To address hypotheses two and three, a two-way, mixed-factor ANOVA was performed on gait parameters, with independent variables of group, task (single or dual-task walking), and their interaction. The main effect of task was used to address hypothesis two, whereas the group × task interaction was used to address hypothesis three. Normal quartile plots were used to detect outliers (three data points from distinct participants were removed prior to analyses based on visual inspection). To address hypothesis four, Pearson bivariate correlation coefficients were used to quantify the associations between each executive function score and each gait parameter within each task. The strengths of these were characterized as strong (0.6-0.8), moderate (0.4-0.6), or weak (0.2-0.4) (Kunter, et al., 2004). JMP 10 (SAS Institute Inc., Cary, NC, USA) was used to carry out statistical analyses, with statistical significance concluded when $p \leq 0.05$.

Results

The first hypothesis was that executive function would be adversely affected by obesity. This hypothesis was supported because the number of words named during the Verbal Fluency letter subtest was 3.3 words lower (i.e. worse) among the obese group (Table 1). Other results that approached statistical significance ($0.05 \leq p \leq 0.10$) were that Stroop Test time was 10.5 s longer (i.e. worse) among the obese group, and Digit Span test was 0.8 points lower (i.e. worse) among the obese group. No effects of age ($p > 0.374$) or years of education ($p > 0.276$) were found for any executive function tests. Scores from executive function tests and gait parameters found here were comparable to other studies. The mean Stroop test completion time of 87.8 s across both groups in the current study was similar, yet slightly shorter, than 100.4 s reported by Jensen (1965) among young adults. The mean Trail Making test completion time of 46.5 s across both groups in the current study was again similar and slightly shorter than the value of ~55 s reported by Tombaugh (2004) among young adults. The median Digit Span score of 2 across both groups in the current study was slightly larger than the median score of ~1.5 reported by Foster et al (1998) among a control group of young adults.

The second hypothesis was that adding a dual-task to walking would increase fall risk among both obese and non-obese participants. This hypothesis was supported because adding a dual-task to walking decreased MTC 0.46 cm, increased stance time 0.08 s, and increased stance time IQR 0.01 s (Table 2). Lower MTC (Aldridge, 2009), higher stance time (Lord, Lloyd, & Li, 1996), and higher stance time IQR (Beauchet, Dubost, Herrmann, & Kressing, 2005; Maki, 1997) have all been associated with a higher tripping/fall risk. Adding a dual-task to walking also reduced RCOF 0.01, which suggests a lower risk of slipping (Wu et al., 2012). The mean MTC in the current study was 1.4 cm across all participants during single-task walking, and was comparable to a mean of ~1.5 cm reported for young adults by Aldridge (2009). The current mean RCOF of 0.21 was also comparable to a mean of 0.20 reported for young adults by Yamaguchi, Yana, Onodera and Hokkirigawa (2013). The mean stance time during single-task walking of 0.67 s was also similar to the mean of 0.65 s for young adults reported by Ortega and Farley (2007).

The third hypothesis was that the increase in fall risk associated with adding a dual-task to walking would be more substantial among obese participants compared to non-obese participants. This hypothesis was not supported because none of the gait parameters exhibited a obesity group × task interaction ($p > 0.327$, data not shown).

The fourth hypothesis was that executive function would be inversely associated with fall risk. This hypothesis was supported based upon two results. First, RCOF during dual-task walking was negatively correlated with the number of words named in Verbal Fluency category subtest (r=-0.32, p=0.044). In other words, a lower Verbal Fluency score was associated with a higher risk of slipping while walking (as indicated by a higher RCOF). Second, stance time IQR during dual-task walking was positively correlated with Stroop Test time (r=0.38, p=0.020). In other words, a lower Stroop Test time was associated with a higher risk of falling (as indicated by a higher stance time IQR). The strength of both correlations was considered weak. No other correlations were statistically significant (data not shown).

Discussion

The current study is the first to our knowledge to demonstrate an association between obesity and executive function among young adults. Lower scores on the Stroop test (12.7% longer completion time), Verbal Fluency (19.8% less words named), and Digit Span test (57.1% greater differences between forward and backward subtest) among obese participants indicated less effective selective attention, semantic memory and working memory. Prior studies that reported adverse effects of obesity on executive function among children used the Stroop test for selective attention, and the Wide Range Assessment of Learning and Memory (WRAML) for working memory (Maayan, Hoogendoorn, Victoria & Antonio, 2011). Obese children named 16.0% fewer number of correct words per minute on the Stroop test, and performed 11.0% worse on the WRAML working memory index (Maayan et al., 2011). The percent difference for the Stroop test was comparable to that reported here.

The physiological mechanism by which obesity adversely affects executive function has not been determined, but obesity has been shown to affect brain morphology. In particular, obesity is associated with smaller cerebellar volume among children (Miller, Couch, Schwenk, Long, Towler & Theriague, et al., 2009), atrophy of the cerebral temporal lobe among older women (Gustafson, Lissener, Bengtsson, Bjorkelund & Skoog, 2004), and atrophy of the frontal, temporal, and subcortical brain regions among older adults (Raji, Ho, Parikshak, Beck & Lopez, 2010). Moreover, smaller grey matter volume was found to be associated with worse semantic memory (Taki, Kinomura, Sato, Goto, Wu & Kawashima, et al., 2011). Willette and Kapogiannis (2014) recently reported a positive association between body fat and frontal grey matter atrophy. Obesity may contribute to compromised cerebellar development and cerebral cortex atrophy through an abnormal accumulation of lipids in the brain (Sriram, Benkovic, Miller & O'Callaghan, 2002), or a free fatty acid excess resulting in a pathological lipid metabolism in the brain (Whitmer, Gunderson, Barrett-Connor, Quesenberry & Jr, Yaffe, 2005).

Changes in gait due to adding a dual-task is thought to result from a competition for attention resources between the two tasks (Woollacott & Shumway-Cook, 2002). In the current study, introducing a dual-task to walking increased fall risk based upon a 33% decrease in MTC, 11.8% increase in stance time, and 33% increase in stance variability. Aldridge (2009) also reported a decrease (albeit only 4%) in MTC when adding talking on a phone to walking while young adults walked on a treadmill at a self-selected speed. Similarly, Beauchet et al (2005) reported an increase (albeit only 6%) in stance time variability when adding backward counting while young non-obese adults walked at a self-selected speed. The substantially larger effect sizes reported here may be due to 1) differences in participant characteristics between studies, such as the current study including half obese participants whereas prior studies included all healthy young adults (Aldridge, 2009; Beauchet et al., 2005); 2) the dual-task used here requiring more attention and thus drawing more attention away from gait; 3) a different method to calculate MTC (the current study used the minimum distance between toe marker and the ground during the mid-swing phase, whereas Aldridge (2009) used the minimum distance between the marker on the 5th metatarsal and the surface of the treadmill). The increase in stance time, although not investigated elsewhere during dual-task walking, may suggest a more cautious walking strategy while simultaneously performing the dual-tasks, and is associated with increased fall risk (Karwowski, 2006), alterations in sensorimotor function (Maki, 1997; Taylor et al., 2013), and deficits in balance control (Karwowski, 2006).

Adding a dual-task to walking did not adversely affect obese participants more than non-obese participants as indicated by a non-significant group × task interaction. This was unexpected given the adverse effect of obesity on executive function, and the increased demands of dual-tasking walking on executive function. One potential reason for this could be that the walking while talking task was not challenging enough to reveal differential effects between the two obesity groups. Another reason could be an insufficient sample size. Future work should consider incorporating different levels of task difficulty in the dual-task condition with a larger sample size.

Similar to the current study, other studies have also reported correlations between the executive function scores and gait during the dual-task conditions. Holtzer, Verghese, Xue, and Lipton (2006) reported executive function (measured by a set of neuropsychological tests including the Digit Span test, Trail Making tests and Verbal Fluency tests, and yielded three neuropsychological factors after factor analysis) to explain 15% of the variance in gait speed during dual-task walking among adults over the age of 70 (obesity status not reported). In addition, Springer et al. (2006) reported an association ($r = 0.61$) between Stroop test score and swing time variability during dual-task walking among older adults age between the ages of 65 and 85with impaired balance. The correlations found in the current study were comparable in strength to the 15% variance reported by Holtzer et al. (2006), but smaller than that reported by Springer et al. (2006), and may be related to the difference in participate age. As a whole, the current study and prior work indicates that gait requires a higher-level control of executive processing, attention, and memory, even among young adults, and that deficits in executive function in people with obesity can contribute to changes in gait that increase fall risk during dual-task conditions.

To the clinician, these results provide evidence that obesity can adversely affect executive function among young adults, and that these adverse effects can, in turn, contribute to changes in gait and risk of falling. It is difficult to identify intervention strategies to mitigate the adverse effects of obesity on executive function because of the limited understanding of the mechanisms by which obesity affects executive function. However, weight-loss may have the potential to improve executive function (Siervo, Arnold, Wells, Taqliabue, Colanbuoni & Albanese et al., 2011). Siervo et al. (2011) reviewed twelve studies which investigated the effects of weight-loss on memory and attention domains of executive function, and performed a meta-analysis and found an improvement in executive function with weight-loss. Resistance training may also have the capability to improve executive functions (Liu-Ambrose, Davis, Nagamatsu, Hsu, Katarynych & Khan, 2010). Liu-Ambrose et al. (2010) found a positive effect of resistance training on executive function (divided attention, working memory, selective attention and response inhibition) among 135 senior women aged 65 to 75 years old who completed a 12-month resistance training.

Five limitations warrant mention. First, although participants were instructed to pay equal attention to both tasks, the amount of attention devoted to each task may have varied across participants or between groups. Second, fall risk was measured indirectly using gait characteristics. Although using gait characteristics to predict fall risk is common, a direct quantitative relationship between these measures and fall risk is not well established. Third, the distribution of body fat was not measured or controlled, but could affect risk of falling (Hilta-Contrearas, Martinez-Amat, Lomas-Vega, Alvarez & Mendoza et al., 2013) and other measures besides BMI used here may better associate with fall risk. Fourth, MTC was evaluated without any obstacle present on the walkway, as is common in the literature. The presence of an obstacle may elicit changes in gait characteristics if seen by the participant. Fifth, as with any cross sectional study, other differences between groups besides the characteristics reported here could have contributed to the results.

The results from the current study provide evidence for three conclusions regarding young adults. First, obesity adversely affected assessments of executive function associated with selective attention, semantic memory, and working memory. Second, adding a talking task to walking increased the risk of tripping, and negatively affected balance control. Third, executive function (i.e. selective attention and semantic memory) was weakly correlated with gait measures associated with risk of falling during dual-task walking, suggesting altered executive function secondary to obesity may contribute to a higher risk of falling.

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Table 1. Group differences in tests of executive function, with summary results presented as means (standard deviation), *F*, *p* and Cohen's *d* values for the main effects of group.

	Obese	Non-obese	*F*	*p*-value	Cohen's *d*
Stroop Test time (s)	93.2 (17.0)	82.7 (13.8)	1.92	0.063	0.65
Trail Making Test time (s)	48.8 (12.5)	44.2 (11.6)	1.12	0.27	0.38
Verbal Fluency (letter)	16.7 (5.0)	20.0 (4.2)	4.35	0.045*	0.66
Verbal Fluency (category)	16.6 (4.2)	15.8 (3.7)	0.63	0.43	0.21
Digit Span	2.2 (1.3)	1.4 (1.4)	3.22	0.081	0.55

Notes: * indicates a statistical difference ($p \leq 0.05$) between groups. Better performance was associated with shorter Stroop Test time, shorter Trail Making Test time, more words in Verbal Fluency Test, and lower score in Digit Span Test.

Table 2. Task differences in gait parameters, with summary results presented as means (standard deviation), and *p*-values for the main effect of task.

	Single-task	Dual-task	Range (Single/Dual)	*F*	*p*-value	Cohen's *d*
MTC (cm)	1.42(0.85)	0.94(0.62)	3.09/2.03	18.66	<0.001*	0.62
RCOF	0.21(0.03)	0.20(0.03)	0.10/0.11	4.53	0.040*	0.19
Stance time (s)	0.68(0.06)	0.76(0.10)	0.47/0.47	68.39	<0.001*	0.89
Stance time IQR (s)	0.04(0.02)	0.05(0.03)	0.07/0.14	7.04	0.012*	0.48

Notes: * indicates $p \leq 0.05$ between groups. MTC=minimum toe clearance; RCOF=required coefficient of friction. Smaller MTC, lower RCOF, shorter stance time and stance time IQR are associated with lower risk of falling. Stance time and stance time IQR were analyzed using raw (without normalization) and after normalizing by leg length. Both analyses yielded similar statistical significance, and the values reported here are the raw values.

Impaired Plantar Sensitivity among the Obese is Associated with Increased Postural Sway

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Abstract

Impaired foot plantar sensitivity has been hypothesized among individuals who are obese, and may contribute to their impaired balanced during quiet standing. The objective of this study was to investigate the effects of obesity on plantar sensitivity, and explore the relationship between plantar sensitivity and balance during quiet standing. Thirty-nine young adults from the university population participated in the study including 19 obese and 20 non-obese adults. Plantar sensitivity was measured as the force threshold at which an increasing force applied to the plantar surface of the foot was first perceived, and the force threshold at which a decreasing force was last perceived. Measurements were obtained while standing, and at two locations on the plantar surface of the dominant foot. Postural sway during quiet standing was then measured under three different sensory conditions. Results indicated less sensitive plantar sensitivity and increased postural sway among the obese, and statistically significant correlations between plantar sensitivity and postural sway that were characterized as weak to moderate in strength. As such, impaired plantar sensitivity among individuals who are obese may be a mechanism by which obesity degrades standing balance among these individuals.

Introduction

An estimated 500 million people worldwide were obese in 2008, and the prevalence of obesity has nearly doubled since 1980 [1]. One of the concerns with the high prevalence of obesity is its association with an increased risk of falls. Each year, obese adults fall almost twice as frequently (27%) as their non-obese counterparts (15%) [2]. This is problematic because falls can be injurious [3]. The biomechanical and/or physiological mechanisms leading to the higher rate of falls among the obese are unclear. Understanding these mechanisms could lead to more effective fall prevention programs.

One mechanism by which obesity could contribute to falls is by degrading balance due to impaired plantar sensitivity on the bottom of the feet. Human standing balance control relies on feedback from the proprioceptive system [4]. This system includes cutaneous mechanoreceptors which detect pressure and deformation in the skin [5]. Studies have demonstrated that impairments in plantar sensitivity influence balance control among older adults and individuals with chronicle ankle instability [6, 7, 8]. Obesity increases postural sway during quiet standing [9], and may do so, at least in part, due to impaired plantar sensitivity. Higher plantar pressures have been reported among individuals who are obese [10], but no studies to our knowledge have investigated the effect of obesity on plantar sensitivity, or the association between plantar sensitivity and balance as a mechanism by which individuals who are obese exhibit impaired balance.

The objective of this study was to investigate the effect of obesity on plantar sensitivity, and explore the relationship between plantar sensitivity and postural sway. Our first hypothesis was that obesity would adversely affect plantar sensitivity. Our second hypothesis was that plantar sensitivity would be associated with postural sway. The results from this study will provide insight to the mechanisms by which obesity impairs balance, and potentially guide future efforts aimed at developing interventions to mitigate the delirious effects of obesity on balance.

Materials and Methods

Thirty-nine young (age=21.3±2.6 years) adults recruited from the university population participated in the study. Participants included 19 obese (body mass index or BMI = 33.0±2.9kg/m^2; 14 females and 5 males) and 20 non-obese (BMI = 22.2±2.2 kg/m^2; 14 females and 6 males) adults. Body fat percentage was also measured using skinfold caliper measurements at the front of the upper arm, back of the upper arm, below the scapula, and on the abdomen (1cm to the right of the navel). Obese participants were required to have a body fat percentage above 35% for women and above 25% for men from these caliper measurements [11], as well as a BMI above 30 kg/m^2. All participants were free from any self-reported foot pain or known neurological conditions that might affect their performance in this test.

Participants completed one experimental session during which multiple measurements of plantar sensitivity were obtained while standing. Plantar sensitivity was operationalized as the force threshold at which an increasing force applied to the plantar surface was first perceived, and the force threshold at which a decreasing force was last perceived. Measurements were obtained immediately upon standing, and at two locations on the plantar surface of the dominant foot including the calcaneus and the head of the third metatarsal. Postural sway was then evaluated under three different conditions. Participants wore a T-shirt, tight-fitting pants, no shoes, and no socks during testing. Room temperature was controlled at 74ºF.

The setup and methodology was based upon a recent study investigating the effects of added weight on plantar sensitivity while upright standing [12]. Plantar sensitivity was assessed using a custom-designed platform (Figure 1) and a digital force gauge (Extech, Model 475040, Nashua, NH, USA. The aluminum platform (40 × 81 cm) was covered in vinyl floor tile and included a 1.5mm diameter hole so that a small stainless steel probe tip (diameter = 1mm) attached to the force gauge could pass through and come into contact with participants' foot sole while standing. The position of the probe tip was controlled from beneath the platform via a manual lab jack (LJ750, Thorlabs Inc, Newton, New Jersey, USA) (Figure 1).

Two practice trials were performed on each participant at the beginning of the experiment. Practice trials were performed at a random site on the plantar surface of the foot not including the two testing sites. Participants were then asked to sit for 10 minutes. To start testing, participants were asked to stand on the platform while the investigator positioned their foot so that the testing site was aligned with the hole in the platform. Participants were instructed to stand as still as possible, look straight forward, hold onto the bars in front of them to help stand still, and give verbal indication when they were able to feel the force by saying "Now". At the start of each trial, the probe tip was initially below the surface of the platform and not in contact with the plantar surface of the foot. After a random delay of up to 10 seconds, the investigator began manually rotating a dial on the lab jack (Figure 1) in increments of approximately 60 degrees every half second until given a verbal indication by the participant. Once the probe tip translated upward far enough to contact the plantar surface of the foot, this rotating pattern increased the force applied to the foot in a step-wise manner at a rate of ~5 grams every half second. After the participant detected the force, this force threshold was recorded, and the investigator continued to raise the probe tip until the force reached 180 grams (a value well above all participants' force threshold). The lab jack was then used to translate the force probe tip downward, resulting in the force applied to the foot decreasing in a step-wise manner at a rate of ~5 grams every half second. Participants were instructed to give verbal indication when they were no longer able to feel the force by saying "Now". A total of four trials were performed at each site, with each trial involving the force increasing and decreasing one time. The order of the two sites was counterbalanced within each group. The experimenter also randomly picked one of the four increasing trials to check whether the participant was giving false verbal indication on their foot sensitivity, or experiencing phantom sensation, by delaying the initiation of the trial for 30 seconds after indicating the start of the trial. None of the participants gave indication before the start of the trial during the experiment.

Postural sway was then evaluated while participants attempted to stand as still as possible with bare feet, arms at sides and feet pointed forward and 7.5cm apart. The trials were collected under three different sensory conditions: eyes-open (baseline), eyes-closed (impaired visual feedback), and eyes-closed with the head tilted backward (impaired visual, vestibular, and proprioceptive feedback). Tilting the head backward is thought to render balance-related vestibular information unreliable by placing the otolith organs outside their normal working range [13]. These conditions were imposed because impairing the visual and vestibular systems would make the balance control system more dependent upon the proprioceptive system (e.g. plantar sensitivity), and may strengthen the relationship between plantar sensitivity and balance. Participants were required to tilt their head backwards at least 30º, as measured by investigator observation. Three trials of 75 s were collected under each condition, and two minutes of rest were allowed in between consecutive trials. The order of the trials was randomized within each group.

During standing trials, ground reaction forces were sampled at 1000 Hz using a force platform (Bertec Corporation, Columbus, OH, USA), and low-pass filtered at 7 Hz using a 4th order Butterworth zero-lag filter . Dependent variables during standing included center of pressure (COP) mean velocity, and COP root mean square (RMS) distance from the mean position in the radial direction. Mean velocity was defined as the total COP distance traveled divided by collection time. The RMS distance was defined as the standard deviation about the mean COP position in radial direction [14]. For each standing trial, the initial and final 5 seconds of the data was removed to avoid initial transients and termination anticipation effects, respectively [14].

A three-way mixed-model analysis of variance (ANOVA) was performed on force threshold measurements with independent variables including group (obese or non-obese), location (head of the third metatarsal or calcaneus) and force direction (increasing or decreasing). A two-way mixed-model ANOVA was performed on postural sway measures with independent variables including group (obese or non-obese) and condition (baseline, impaired vision, or impaired vestibular feedback). Tukey's Honestly Significant Difference procedure was used to investigate pair-wise comparisons of interest in the event of significant interactions. A log transform on force threshold measurements and postural sway measures was performed prior to the analyses to achieve a normal distribution of residuals.

Bivariate correlation analysis was performed between plantar sensitivity and postural sway measures, and the strength of the correlation was quantified using the Pearson product-moment correlation coefficient. Mean of the four plantar sensitivity trials under each testing location and force direction was correlated with the mean of the three postural sway trials under each condition. Two data points were associated with substantially larger Cook's distance values [15], and thus were excluded from the bivariate analyses to avoid a disproportionate influence on the correlation. The strength of correlations were characterized using the correlation coefficient (r) as strong (0.6-0.8), moderate (0.4-0.6), and weak (0.2-0.4) [16]. JMP 10 (SAS Institute Inc., Cary, NC, USA) was used to carry out the statistical analyses, and statistical significance was concluded if $p \leqslant 0.05$.

Results

Force threshold measurements exhibited a mean value of 29.8 grams and a range of 2 - 156 grams across all groups (Figure 2). It exhibited a group by location by force direction interaction (p=0.040; Figure 3). Under the third metatarsal, the obese group exhibited a 56% higher force threshold when force was increasing (p<0.001), and a 22% higher force threshold when force was decreasing (p=0.008). Under the calcaneus, the obese group exhibited a 30% higher force threshold when force was increasing (p=0.019), and a 22% higher force threshold when force was decreasing (p<0.001). The force threshold across both groups was 79% higher at the calcaneus compared to the third metatarsal when force was increasing (p<0.001), but not significantly different between these locations when force was decreasing. Lastly, the force threshold across both groups was 183% higher under the calcaneus when force was increasing (p<0.001), and 91% higher under the third metatarsal when force was increasing (p<0.001).

Both mean velocity and RMS distance exhibited no group by condition interaction (p=0.344 for mean velocity and p=0.179 for RMS distance), but did exhibit main effects of group (p=0.008 for mean velocity and p=0.016 for RMS distance) and condition (p<0.001 for mean velocity and RMS distance). The obese group exhibited 5% higher mean velocity and 7% higher RMS distance compared to the non-obese group.

Bivariate analyses revealed positive correlations of weak to moderate strength between two plantar sensitivity measurements (calcaneus/decreasing force and 3rd metatarsal/increasing force) and both postural sway measurements under the three different conditions (Figure 4). No other correlations were significant.

Discussion

The objective of this study was to investigate the effect of obesity on plantar sensitivity, and explore the relationship between plantar sensitivity and postural sway. We hypothesized obesity would adversely affect plantar sensitivity. The threshold to detect force onset and offset were both higher among obese participants at both locations tested. These results indicated obese participants were unable to sense plantar forces as small as non-obese participants while standing. As such, this hypothesis was accepted. We also hypothesized that plantar sensitivity would be associated with postural sway. This hypothesis was accepted because weak to moderate correlations were found between the postural sway and plantar sensitivity measures.

The mean force thresholds, and the effects of obesity on force thresholds, measured here are comparable to a prior study that investigated the effects of adding body mass on plantar sensitivity [12]. Handrigan et al. [12] measured the minimum force threshold while standing, and compared these measurements before and after adding 23 kg of body mass with a weighted vest.

This added body mass corresponded to an increase in BMI from 24.7 to 31.9 kg/m^2. The mean force threshold reported by Handrigan et al. [12] was 20.0 g before adding weight, which was comparable to the 25.6 g threshold measured among non-obese participants in the current study across all conditions. Handrigan et al. [12] also reported a 30% increase in force threshold after adding 23 kg of body mass, which was comparable to the 29.3% higher threshold among the obese group in the current study across all conditions. Together, these studies provide evidence for the added body mass associated with obesity eliciting impaired plantar sensitivity.

Two characteristics of mechanoreceptors may help to explain the impaired plantar sensitivity among obese participants found here. First, the relationship between mechanical stimulus intensity and internal mechanoreceptor voltage potential is nonlinear, and suggests less sensitivity at higher intensities [17]. At lower stimulus intensities, only slight changes in stimulus intensity are needed to markedly increase the mechanoreceptor potential. At higher stimulus intensities, however, an equivalent increase in stimulus intensity only results in a slight increase in mechanoreceptor potential. The higher plantar pressure associated with obesity [10] may result in the operating range of the plantar mechanoreceptor to be closer to the range over which changes in stimulus intensity only result in slight changes in mechanoreceptor potential. Second, the Weber-Fechner Principle states that the amount of change in mechanical stimulus intensity necessary for detection is proportional to the stimulus intensity [18]. As such, the higher plantar pressure associated with obesity would require a larger change in plantar mechanical stimulus to be detected, and result in reduced sensitivity on the plantar surface of the foot.

Force thresholds were lower when force was decreasing than increasing. We have two possible explanations for these results. First, Meissner's corpuscles and Pacinian corpuscles are two rapidly adapting mechanoreceptors responsible for detecting the onset and offset of a mechanical stimulus [19]. Meissner's corpuscles are thought to be more amenable to detecting light touch due to their relatively superficial distribution within the skin, and Pacinian corpuscles are thought to be more amenable to detecting deep pressure due to their deeper distribution within the skin [19]. Given these differences, detecting the onset of an initially zero force that increased may be mostly dependent upon Meissner's corpuscles, whereas detecting the offset of an initially high force (that stimulated deeper Pacinian corpuscles) and decreased may have involved both Meissner's and Pacinian corpuscles. Involvement of Pacinian corpuscles may have offered more sensitivity while force was decreasing because Pacinian corpuscles have a lower response threshold than Meissner's corpuscles [20]. Our second possible explanation is related to the mechanical properties of the skin [21]. Skin is viscoelastic, which means its mechanical response to force depends upon time. Measurements as force increased were made shortly (less than 18 seconds, on average) after the force was applied, and measurements as force decreased were made after the force was applied for at least 30 seconds. Therefore, the viscoelastic stress relaxation that would have occurred during testing likely resulted in differences in the mechanical state (i.e. tissue strain) at the testing site that may have contributed to differences in force thresholds.

Our results indicated that plantar sensitivity is weakly to moderately correlated with postural sway. Mechanoreceptors are preferentially distributed in the anterior, lateral border and heel regions of the plantar surface [20], which could correspond to the critical regions of the foot that support the majority weight of the body under the weight-bearing condition [22]. Similarly, mechanoreceptors under the anterior and posterior regions of the feet provide feedback that regulate the body to tilt posteriorly and anteriorly [22]. The central nervous system may be able to extract a spatial distribution cue according to the plantar pressure which could be transformed into a body position cue indicating the direction and the amplitude of the whole-body inclination [23]. Therefore, deficits in plantar sensitivity could have a direct influence on balance control [6, 7, 8]. However, the mechanism of why only calcaneus/decreasing force and 3rd metatarsal/increasing force were correlated with postural sway measurements under the three different conditions is not clear.

The postural sway condition with eyes closed and head tilted backward was included based upon the expectation that any impairment in plantar sensitivity among obese participants would have a greater effect on postural sway when input from the visual and vestibular systems were eliminated or altered. Based upon the lack of a group by condition interaction in sway measures, this expectation was not met. This may have been due to not tilting participants head backward sufficiently far, or input from other components of the proprioceptive system offsetting the altered input from the plantar surface among the obese. Nevertheless, postural sway exhibited weak to moderate correlation with plantar sensitivity despite not adequately impairing vestibular input during sway measurements.

Several limitations are to be noted for this study. First, changes in force were stepwise and in increments of approximate 5 gram per half second, rather than at a steady rate. This may have limited our force threshold resolution 5 grams, but this amount of force was ostensibly sufficiently small to identify the effects of interest. Second, the results may be dependent upon the nature and duration of activities performed before testing (someone may have been standing for hours while another may have been laying down). However, these data were not obtained. Third, plantar sensitivity was only obtained from the dominant foot, and it is unclear if any left/right asymmetry in plantar sensitivity is common. Fourth, the amplitude of head tilting was not strictly controlled in the current study, and may have contributed to some inter-subject variability in the vestibular feedback during this condition.

In conclusion, obesity impaired plantar sensitivity among a cohort of young adults. This impairment was associated with increased postural sway during quiet standing, and may be a contributing factor to the increased fall risks among individuals who are obese.

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Conflict of Interest
None

Reference

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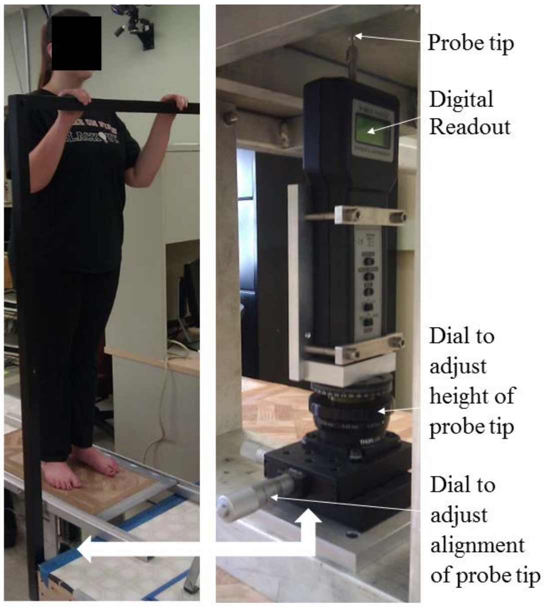

Figure 1 Custom-designed platform to assess plantar sensitivity while standing (Digital force gauge setup was mounted on lab jack under the platform. The investigators adjusted the vertical position of the probe by manipulating the lab jack, and read off the force threshold from the digital readout when indicated by the participants.)

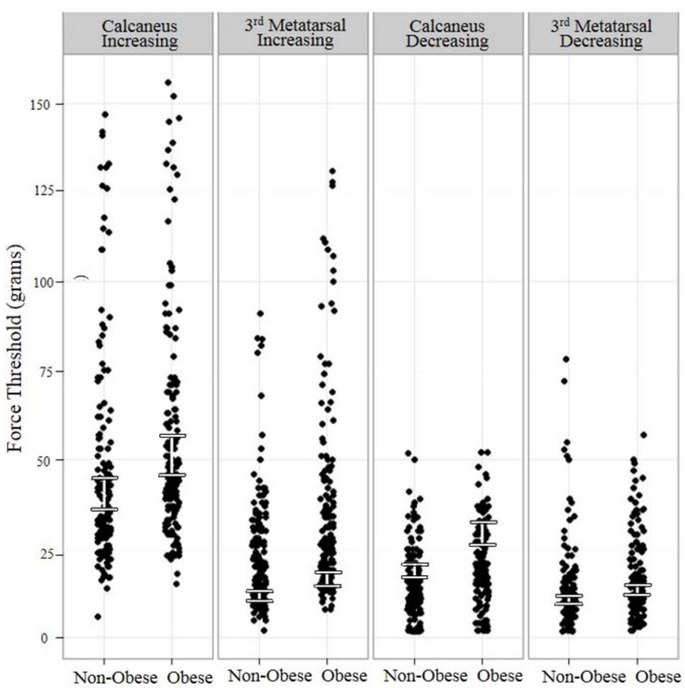

Figure 2 Force threshold measurements separated by group, location, and force direction. Dots indicate individual measurements, and brackets indicate 95% confidence intervals of the mean.

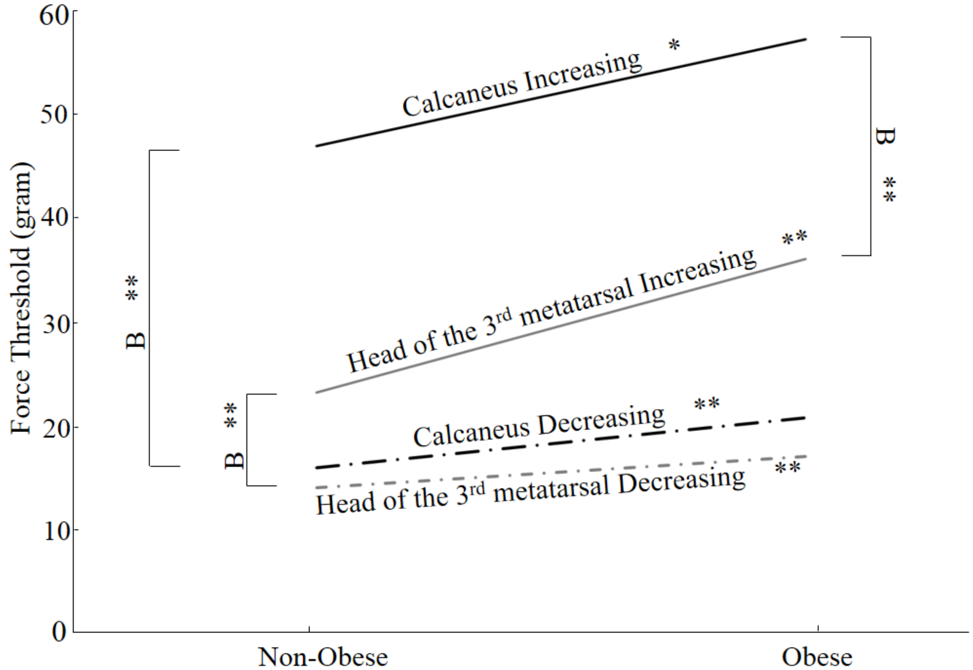

Figure 3 Group by location by force direction interaction plot illustrating differences between groups and conditions * indicates $p \leq 0.05$, **indicates $p \leq 0.01$. B indicates statistical significance when combining across both groups.

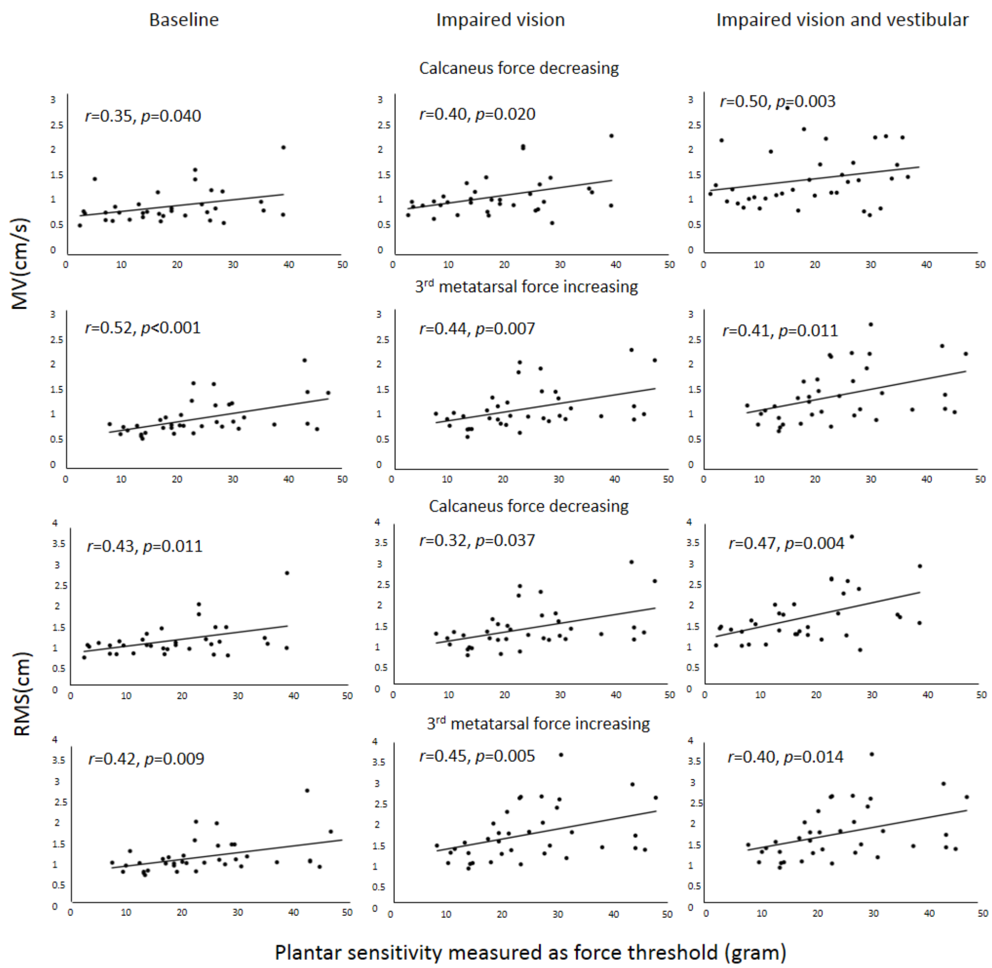

Figure 4 Scatter plots of correlation between plantar sensitivity (calcaneus/immediately upon standing/decreasing force and 3rd metatarsal/immediately upon standing/increasing force) and postural sway measures (mean velocity and RMS distance) with *r* and *p* values.

Falls resulting from a laboratory-induced slip occur at a higher rate among individuals who are obese

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Abstract
Falls due to slipping are a serious concern, with slipping estimated to cause 40-50% of all fall-related injuries. Epidemiological data indicates that older and obese adults experience more falls than young, non-obese individuals. An increasingly heavier and older U.S. population and workforce may be exacerbating the problem of slip-induced falls. The purpose of this study was to investigate the effects of obesity and age on slip severity and rate of falling resulting from a laboratory-induced slip. Four groups of participants (young obese, young non-obese, older obese, older non-obese) were slipped while walking at a self-selected, slightly hurried pace. Slip severity (slip distance, slip duration, mean slip speed and peak slip speed) and slip outcome (fall or recovery) were compared between groups. Obese individuals experienced 22% faster slips than non-obese individuals in terms of mean slip speed (p=0.022). Obesity did not affect slip distance, slip duration or peak slip speed. Obese individuals also exhibited a higher rate of falls; 32% of obese individuals fell compared to 10% of non-obese (p=0.005). Obese individuals were more than eight times more likely to experience a fall than non-obese individuals when adjusting for age, gender and gait speed. No age effects were found for slip severity or slip outcome. These results, along with epidemiological data reporting higher fall rates among the obese, indicate that obesity may be a significant risk factor for experiencing slip-induced falls. Slip severity thresholds were also reported that may have value in developing controls for fall prevention.

1. Introduction
Occupational falls and fall-related injuries are a major source of mortality, morbidity, and medical expense. In 2011, falls accounted for 553 occupational fatalities and 22% of injuries requiring days away from work in the U.S. (Stephen and Janocha, 2013). The annual cost of occupational fall-related injuries is estimated to be $5.7 billion in the United States alone (Yoon and Lockhart, 2006). Slipping causes an estimated 40-50% of all fall-related injuries (Courtney et al., 2001). For example, slips caused 85% of falls to the floor, and 30% of falls to a lower floor during construction of the Denver International Airport between 1989 and 1994 (Lipscomb et al., 2006).

The problem of occupational slip-related falls may be exacerbated by the high prevalence of obesity. In the 1960s, only 13% of the U.S. population was considered obese based upon body mass index, or BMI (Wang and Beydoun, 2007). In 2011-2012, 69% of the population was estimated to be overweight ($25 < \text{BMI} < 30$ kg/m^2) or obese (BMI ≥ 30 kg/m^2), and 35% was estimated to be obese (Ogden et al., 2012). Fall rates are higher among obese individuals (Fjeldstad et al., 2008; Himes and Reynolds, 2012; Patino et al., 2010). These higher fall rates could be due to an impaired ability to maintain balance after slipping, secondary to the adverse biomechanical and physiological effects of obesity (Madigan et al., 2014). For example, muscle strength relative to body weight is lower among obese compared to non-obese individuals

(Lerner et al., 2014), and could reduce net muscle moments and movement capabilities when trying to maintain balance after slipping. Trunk segment inertia is higher among obese individuals (Matrangola et al., 2008), which would increase the trunk's kinetic energy during walking, and require additional joint work to attenuate when attempting to maintain balance after slipping. Lastly, plantar sensitivity is degraded among obese individuals (Wu and Madigan, 2014), and could contribute to delayed slip onset detection. This would lead to longer and faster slips that require higher net muscle moments to maintain balance.

The problem of slip-related falls in the workplace could also be exacerbated as more adults are continuing to work later in life because fall rate increase with age (Tromp et al., 2001). Moreover, individuals who are obese *and* older would seem to be at a particularly elevated risk for falling due to the combination of risk factors associated with obesity and aging. However, the combined effects of obesity and age on slips and falls has not been investigated. In a letter to the editor, Zhang et al. (2014) reported older adults who were obese were 2.5 times more likely to fall while walking following a simulated slip using a sliding platform (Zhang et al., 2014). This study provides evidence that obese older adults are less able to maintain balance after a large postural perturbation while walking compared to non-obese older adults. Given kinematic differences between these simulated slips and actual slips (Troy and Grabiner, 2006), these results may not generalize to actual slips on a low-friction surface. It would also be beneficial to explore slipping kinematics in order to begin to understand the underlying reason(s) for the hypothesized higher rate of slip-induced falls among obese individuals.

The purpose of this study was to investigate the effects of obesity and age on slip severity and the rate of falling after a laboratory-induced slip. Slip severity was quantified using the motion of the slipping foot during the slip, and was operationalized by slip duration, slip distance, peak slip speed, and mean slip speed. These measures were selected based upon their use in other slip studies, and their association with fall outcome after slipping (Brady et al., 2000, Troy et al., 2008, Yang et al., 2012). Three hypotheses were tested: (1) obese adults would experience more severe (i.e. longer in duration, farther, and faster) slips than non-obese adults; (2) obese adults would fall at a higher rate after slipping; (3) an obesity × age interaction would exist, in which the effects of obesity on slip severity and rate of falling would be magnified among older than younger adults. These hypotheses were based upon the potential for previously reported biomechanical and physiological factors impairing the ability to maintain balance after slipping. Any effects of obesity or age on slip severity or rate of falls can be used to help identify workers at an increased risk of falling from a slip, and potentially guide subsequent investigations aimed at developing fall prevention strategies.

2. Methods

Seventy-two adults completed the study and were categorized into one of two age groups and one of two obesity groups. Participants included 26 young (18-29 years) non-obese (BMI 17.7-24.9 kg/m^2); 25 young obese (BMI 29.1–40.4 kg/m^2); 10 older (50-66 years) non-obese (BMI 19.5-26.3 kg/m^2); and 11 older obese (BMI 30.1–45.1 kg/m^2) individuals. Participants were recruited from the university and local community using electronic announcements, community flyers, and newspaper advertisements. All participants were screened using a medical questionnaire to exclude individuals with self-reported musculoskeletal or neurological disorders that could affect their gait or balance. Participants aged 65 and older, as well as one 60-year-old participant based upon responses to the medical questionnaire, were also required to pass a medical evaluation administered by a physician. This evaluation excluded participants with any neurological, cardiac, respiratory, ontological, or musculoskeletal disorders, and required a minimum bone density of the femoral neck of 0.65 g/cm^2 as assessed by DXA (General Electric, Lunar Digital Prodigy Advance, Madison, WI). The university Institutional Review Board approved this study, and informed consent was obtained from all participants prior to participation.

The experiment involved one experimental session during which participants were exposed to an unexpected slip while walking on a level walkway. Participants were asked to look straight ahead and walk at a self-selected, yet purposeful (slightly hurried), speed along a 10-meter walkway covered in vinyl tile. Initially, participants performed five to 10 walking trials to acclimate to the lab environment, and to establish a starting position so that the dominant foot (preferred foot to kick a ball) naturally and consistently landed on a force platform integrated in the middle of the walkway. Participants were then informed that a slip or trip may or may not occur during any subsequent walk down the walkway, and that if slipped or tripped, they should attempt to maintain their balance and continue walking. All participants wore shoes with polyvinylchloride soles to prevent any frictional differences between participants at the shoe-floor interface. To prevent impact with the floor in the event of an impending fall, participants wore a safety harness connected with a lanyard to a track above the walkway. The length of the lanyard was adjusted so that a participant's hands could not touch the floor and their knees would be 15 cm from the floor when kneeling.

To divert attention from walking and the possibility of a slip or trip, participants watched television (one positioned at each end of the walkway) and listened to the audio through wireless headphones while walking. For approximately 1-2 minutes between consecutive trials, participants stood at the end of the walkway, with their backs to the walkway, and watched the television until instructed to turn around and begin the next trial. This time interval allowed the investigators to prepare for the next trial. After a minimum of 10 acceptable walking trials (with proper foot placement on the force platform and walking/sacrum marker speed not fluctuating by more than ± 0.1 m/s between trials), a foam paint roller was used to apply a uniform layer of vegetable oil (50 mL) to the entire surface of the force platform (0.9 m × 0.9 m) while the participant faced away from the walkway. The next walking trial then continued using the same procedure as earlier trials. A slip occurred when the dominant foot contacted the contaminated surface, and all participants were successfully slipped on the first attempt. The lights in the lab were dimmed throughout testing to reduce any glare/visual cue created by the slip contaminant.

During walking and slipping trials, the three-dimensional position of 28 reflective markers were sampled at 100 Hz using a Vicon motion capture system with six T10 cameras (Vicon Motion Systems, Centennial, CO) and low-pass filtered at 7 Hz (second-order, zero-phase-lag Butterworth filter). Reflective markers were placed at anatomical landmarks based on a modified Helen Hayes marker set (Kadaba et al., 1990). Ground reaction forces under the slipping foot, and the force applied to the harness, were sampled at 1000 Hz using a 6-degree of freedom force platform (Bertec Corporation, Columbus, OH) and a uniaxial load cell (Cooper Instruments and Systems, Warrenton, VA), respectively. Both were low-pass filtered at 20 Hz (second-order, zero-phase-lag Butterworth filter).

Six dependent variables were obtained from each slip trial using custom-written code in Matlab 2013a (The Mathworks Inc., Natick, MA): gait speed, gait step length, slip duration, slip distance, peak slip speed, and mean slip speed. Gait speed was calculated as the forward speed of a marker on the sacrum for 2-3 steps preceding slip onset (Brady et al., 2000). Slip severity parameters were calculated by first identifying two events: slip onset and slip end (Brady et al., 2000; Redfern et al., 2001; Strandberg and Lanshammar, 1981). Slip onset was defined as heel contact onto the contaminated surface using force data collected from the force platform (Brady et al., 2000). The slip end was defined as the time when either: (1) the slipping foot slipped off the edge of the force platform, (2) the heel came to a stop, or (3) the heel displaced vertically from the force platform (Brady et al., 2000; Cham and Redfern, 2001). Slip distance was calculated as the total distance traveled from slip onset to slip end, and slip duration was calculated as the time from slip onset to slip end. Peak slip speed was calculated as the maximum speed of the heel marker during the slip (Brady et al., 2000) using a finite difference algorithm.

Slip outcome (fall or recovery) was determined using forces collected from the harness load cell and methods described elsewhere (Brady et al., 2000). Slip outcome was classified as: (1) *recovery* if the peak harness load was < 30% body weight (BW) and the integrated harness load was < 8% BW × second; (2) *fall* if the peak harness load was ≥ 50% BW; (3) *harness-assisted* otherwise. Harness-assisted trials were excluded from further analysis due to their ambiguous outcome had participants not been wearing the harness. Slip severity thresholds that correctly separated the highest percentage of falls and recoveries were determined by iteration.

Differences in gait speed and gait step length between obesity groups and between age groups were investigated using separate two-way analyses of variance. The interaction effect (obesity x age) was not statistically significant for gait speed (p=0.166) and gait step length (p=0.147), and was subsequently removed from both analyses. The four slip severity measures (slip duration, slip distance, peak slip speed, and mean slip speed) were analyzed using separate three-way analyses of covariance with independent variables of obesity group, age group, and gender, and with gait speed as a covariate. All three-way and two-way interaction effects were initially included in the analysis. Iterative backwards elimination was then used to remove non-significant three-way and two-way interactions until the final model included only main effects and significant interactions. Following this procedure, no interactions remained in the final models for any of the dependent variables. Slip outcome was analyzed using a logistic regression model with independent variables of obesity group, age group, gender, and gait speed. An obesity group × age group was initially included, but subsequently removed because it was not statistically significant. Gait speed and gait step length were normalized by body height prior to statistical analysis, but reported in units of meters and meters/second for utility. Statistical analyses were performed using JMP 10 (SAS Institute Inc., Cary, NC) with a significance level of $p \leq 0.05$.

3. Results

Gait speed and gait step length were not affected by obesity group (speed p=0.486, step length p=0.886) or age group (speed p=0.245, step length p=0.593; Table 1). One of four slip severity measures was affected by obesity group (Table 1). Mean slip speed was 22% higher (p=0.022) among obese participants, but slip duration (p=0.974), slip distance (p=0.121), and peak slip speed (p=0.065) were not affected by obesity group. Age group did not affect slip severity measures, including slip duration (*p*=0.112), slip distance (*p*=0.933), peak slip speed (*p*=0.591), and mean slip speed (*p*=0.543; Table 1). Gait speed affected three of four slip severity measures. Slip distance (*p*=0.005), peak slip speed (*p*<0.001), and mean slip speed (*p*<0.001) increased as gait speed increased, but gait speed did not affect slip duration (*p*=0.148).

Slip outcome differed between obesity groups (*p*=0.005; Figure 1) in that 10 out of 31 (32%) obese participants fell after slipping compared to 3 out of 30 (10%) non-obese participants. The odds ratio for obesity group indicated that obese participants were 8.24 [95% C.I.: 1.81, 57.10] times more likely to fall than non-obese participants when adjusting for age group, gender, and gait speed. Slip outcome was not affected by age group (*p*=0.937) or gender (*p*=0.395; Table 2). However, a fall was more likely as gait speed increased (*p*=0.003), with an odds ratio of 1.08 [95% C.I.: 1.02, 1.14] for a 1 cm/s increase in gait speed when adjusted for obesity group, age group, and gender.

Inspection of the data revealed that falls were associated with more severe slips. Falls were associated with longer slip distances, but the slip distance for some recovery trials surpassed slip distances resulting in falls (Figures 2 and 3). Similarly, falls were associated with slips that were longer in duration, and faster in speed, but the duration and speed of some recovery trials surpassed some resulting in falls (Figure 3). As such, there were no slip severity thresholds that separated all falls from all recoveries. Participants who fell tended to experience more severe slips. In general, the majority of falls occurred with slip distances beyond 50 cm, slip durations longer than 0.3 s, peak slip velocities above 2.5 m/s, and mean slip velocities over 1.0 m/s. More specifically, a slip distance of 56.5 cm separated 85.4% of recoveries from 86.7% of falls, a slip duration of 0.35 s separated 54.2% of recoveries from 86.7% of falls, a peak slip speed of 2.57 m/s separated 91.7% of recoveries from 80.0% of falls, and a mean slip speed of 1.19 m/s separated 79.2% of recoveries from 86.7% of falls (Figure 3).

4. Discussion

The purpose of this study was to investigate the effects of obesity and age on slip severity and the rate of falling after a laboratory-induced slip. Our first hypothesis was that obese adults would experience more severe slips than non-obese adults. This hypothesis was accepted because mean slip speed was higher among obese adults. Additionally, a post-hoc power analysis indicated peak slip speed would have be statistically significant with only four more participants added to each group (assuming the same standard errors and structural results as the current sample). This relatively small number of additional participants perhaps suggests a greater emphasis on the results than the lack of statistical significance would indicate. It is interesting to note that the range of slip severity parameters for falls and recoveries found here agreed well with those reported by Brady et al., despite experimental differences including the use of mineral oil to slip young adults age 26.6 ± 3.9 years who were walking barefoot.

Slip severity was related to slip outcome in that more falls resulted from more severe slips (Figure 3). However, there were exceptions to this trend. For example, falls occurred after slip distances as short as 35.4 cm, and recoveries occurred after slip distances as long as 72.2 cm. Previously reported slip severity thresholds suggested falls would likely occur when slip distance exceeded 10 cm and peak slipping speed exceeded 50 cm/s (Cham and Redfern, 2002b; Perkins, 1978; Strandberg and Lanshammar, 1981). These previously reported thresholds were substantially smaller than those reported here (slip distance = 56.5 cm and peak slipping speed = 2.57 m/s). The large discrepancies between these studies may be attributed to methodological differences such as these prior studies employing multiple slips, and more recent work documenting a reduction in slip severity with repeated exposure to slipping (Siegmund et al., 2006). These thresholds for slip distance and peak slip speed have potential value in preventing falls from slips. Smooth flooring can be desirable for aesthetics or ease of cleaning. Smooth flooring, however, is typically associated with a higher likelihood of slipping in the presence of a liquid contaminant. Using the slip severity thresholds identified here, smooth flooring could be designed to include high friction markings or boundaries that would arrest a slip. As long as these markings or boundaries were designed to limit any slip below the threshold identified here, they may help prevent falls after slipping. This potential engineering control would need to be confirmed experimentally.

Our second hypothesis was that obese adults would fall at a higher rate after slipping. This hypothesis was accepted because obese adults were over eight times more likely to fall after slipping compared to non-obese adults after adjusting for age group, gender, and speed. Several underlying neuromuscular alterations associated with obesity (Madigan et al., 2014) could have contributed to the higher fall rate and less favorable trip recovery measures with obesity. First, obesity is associated with increased trunk mass (Matrangola et al., 2008), which would increase the mechanical demands necessary to arrest trunk motion when attempting to maintain balance after slipping. Second, obesity is associated with reduced lower extremity strength relative to body mass (Wearing, 2006), which would seem to compromise the capacity to enact a quick and effective stepping response necessary to maintain balance after slipping. Third, sensory deficits could contribute to a delayed response and more severe slip (Wu and Madigan, 2014).

Our third hypothesis was that an obesity × age interaction would exist whereby the effects of obesity on slip severity and rate of falling would be magnified among older adults compared to young adults. This hypothesis was rejected because no obesity group × age group interactions existed for any slip severity measures, or fall rate. Moreover, no age group differences in fall rate and slip severity were found. These results differ from previous studies that report older adults experienced higher fall rates and faster/farther slips following laboratory-induced slips (Troy et al., 2008, Lockhart et al., 2003). Regarding fall rate, Troy et al. (2008) reported that 18 out of 21 (86%) older adults (mean age 71 years) fell, and 2 out of 32 (6%) young adults (mean age 25 years) fell. Lockhart et al. (2003) reported that 7 out of 14 (50%) older adults (mean age 76 years) fell, and 2 out of 14 (14%) young adults (mean age of 23 years) fell. Regarding slip severity, Troy el al. (2008) reported slip distance was 39.8 cm among older fallers and 19.6 cm among young non-fallers, while peak slip velocity was 2.37 m/s among older fallers and 1.40 m/s among young non-fallers. Lockhart et al. (2003) reported initial slip distance was 2.2 cm among older adults and 1.1 cm among young adults, while sliding heel speed was 0.76 m/s among older adults and 0.47 m/s among young adults. Unfortunately, a direct comparison of slip severity parameters between these prior studies and the current study is difficult due to differences in how slip severity parameters were calculated and summarized. In particular, Troy et al. only reported slip severity measures of older fallers and young non-fallers. Lockhart et al. calculated sliding heel speed as the mean speed from slip onset to peak sliding heel speed, initial slip distance as distance from slip onset to peak heel horizontal acceleration, and a fall having occurred when sliding heel speed exceeded the whole-body center of mass speed during the slip. Additionally, older adults in these previous two studies had mean ages of 71 and 76 years, whereas older adults in the present study had a mean age of 58 years. This likely contributed to the differences in age-related effects between the studies.

Another factor influencing fall outcome was gait speed. Faster gait speed was associated with more severe slips and more falls. These results with respect to fall rate do not agree with prior work. Two previous studies used mineral oil (Brady et al., 2000) or a water-detergent mixture (Hu and Qu, 2013) to slip participants walking at a self-selected speed. Brady et al. reported no effect of gait speed on slip outcome, and Hu and Qu found no difference in walking speed between falls and successful recoveries. Other studies have reproduced slips with sliding platforms, and reported faster gait speeds to improve stability and reduced rate of falling (Bhatt et al., 2005; Espy et al., 2010). As such, the influence of gait speed on slip outcome remains an open question.

Several limitations were present in this study. First, participants were warned of a possible slip, so anticipation effects may have existed (Cham and Redfern, 2002a; Heiden et al., 2006; Horak and Nashner, 1986). Second, the age range of our older participants was 50-66 years, and was chosen to correspond with the final years of the traditional working age. As such, our results with respect to age may not generalize to adults over the age of 65. Third, the end of slip for several participants occurred when the foot slipped up to (and over) the edge of the force platform. As such, it was unclear as to how far these participants would have slipped had the force platform been extended further on the walkway. Fourth, it is unclear if slip severity and fall rates are dependent upon the contaminant and shoe sole material employed. Any such dependence would potentially influence the ability to generalize parameter values between slipping studies and outside of the laboratory.

5. Conclusions

Obesity increased slip speed and fall rate after a laboratory-induced slip. These results suggest that the higher fall rate reported among individuals who are obese may be due, at least in part, to a reduced ability to maintain balance after slipping. Slip severity thresholds that separated the majority of falls from recoveries were also reported. These threshold values may have practical value in preventing falls from slips by incorporating friction markings or boundaries to limit slip distance to values lower than when falls occurred. This would need to be confirmed experimentally.

Conflict of interest statement

The authors have no conflicts of interest to disclose.

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Table 1. Gait and slip severity measures (mean ± standard deviation)

	Gait Measures		Slip Severity Measures			
	Gait Speed (m/s)	Step Length (m)	Slip Duration (s)	Slip Distance (cm)	Peak Slip Speed (m/s)	Mean Slip Speed (m/s)
Obese	1.26±0.1	0.70±0.0	0.34±0.1	40.2±28.3	1.83±1.	1.01±0.6
Non-	1.29±0.1	0.70±0.0	0.35±0.1	34.4±22.3	1.58±0.	0.83±0.4
Older	1.31±0.1	0.69±0.0	0.30±0.0	35.0±25.8	1.60±0.	0.97±0.6
Young	1.27±0.1	0.71±0.0	0.36±0.1	38.0±25.5	1.74±0.	0.91±0.5
Male	1.28±0.1	0.72±0.0	0.31±0.0	29.5±21.7	1.50±0.	0.81±0.4
Female	1.27±0.1	0.68±0.0	0.37±0.1	44.1±26.7	1.89±0.	1.01±0.5

* indicates statistical difference ($p \leq 0.05$) between obesity groups. Slip distance, peak slip speed, and mean slip speed all showed a significant increase with gait speed.

Table 2 Slip outcomes among groups

	Fall	Recovery	Harness-Assisted	Total
Total	13	48	11	72
Obese	10*	21	5	36
Non-obese	3*	27	6	36
Older	3	12	6	21
Young	10	36	5	51
Male	4	24	5	33
Female	9	24	6	39

* indicates statistical difference ($p \leq 0.05$) between obesity groups

Logistic regression model to predict probability of a fall:

logit = -10.21 - 0.96OB_GRP - 0.30AGE_GRP - 0.01GEN + 11.17SP

where: AGE_GRP = -1 for young and 1 for older
OB_GRP = -1 for obese and 1 for non-obese
GEN = -1 for males and 1 for females
SP = speed in %BH/s
Probability of fall = $1/(1+e^{-logit})$

Using a cutoff score of 0.5: Model Sensitivity = 25%
Model Specificity = 94%

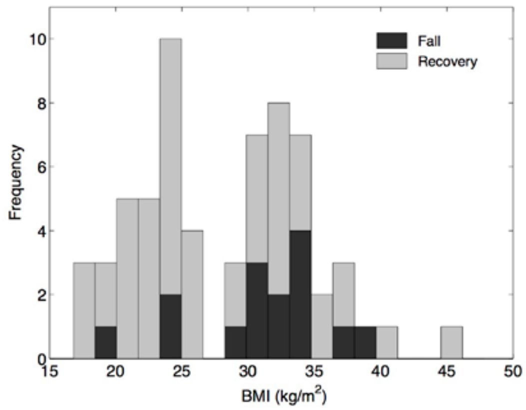

Fig. 1. The distribution of falls and recoveries across BMI shows a higher number of falls among the obese group.

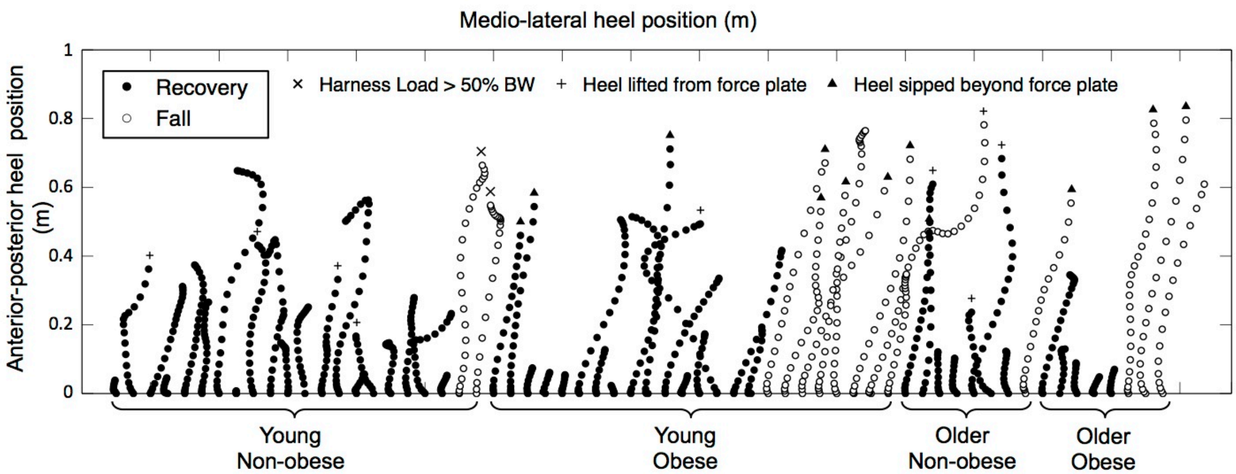

Fig. 2. Sliding heel trajectories for all slips classified as recoveries or falls, organized based on slip distance and age/BMI group. Each data point represents the sliding heel position from slip onset to slip end, separated by 20 ms intervals. Slips ended when the sliding heel came to a stop, unless otherwise indicated.

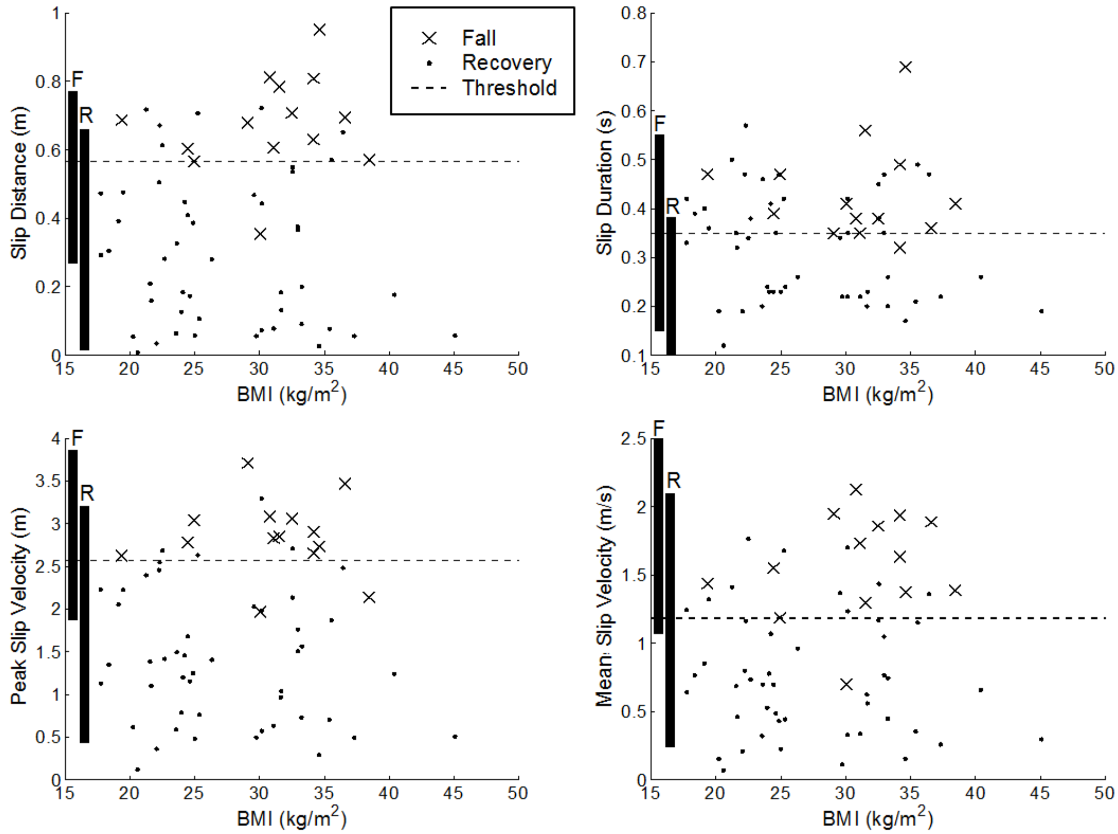

Fig. 3. Slip severity parameters from all slip trials were plotted as a function of subject BMI. Most falls occurred among the obese group, and falls were associated with more severe slips. Dashed slip severity threshold lines separated most falls from most recoveries. Vertical bars on the left of each subplot illustrate the range of slip severity parameters for falls (F) and recoveries (R) resulting from a laboratory-induced slip as reported by Brady et al. (2000).

Falls resulting from a laboratory-induced trip occur at a higher rate among individuals who are obese

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TECHNICAL ABSTRACT
Background: Obese adults are reported to fall at a higher rate than non-obese adults. **Purpose:** To help determine the reason for this higher fall rate, we quantified fall rates and balance recovery measures in two participant groups (obese and non-obese) after a laboratory-induced trip. Our focus was on young adults given that they comprise a substantial portion of the workforce. **Methods:** Twenty-one young adult participants (11 obese, 10 non-obese) walked along a 10 m walkway at a purposeful (slightly hurried) speed. During a randomly selected walking trial, an obstacle was raised to elicit a trip near mid-swing. **Results:** Three participants who fell employed a lowering stepping strategy to recover balance, while 14 out of 16 participants who recovered balance employed an elevating strategy (the remaining participants did not clearly fall or recover). Falls were associated with a significantly longer time to complete the initial recovery step, a larger trunk angle, and angular velocity ($p = 0.012$) at foot strike of the first recovery step, and a lower hip height after stepping. Fall rates were notably, but not significantly, higher among obese participants (30%) compared to non-obese participants (0%). The time to complete the initial recovery step was longer among obese participants, but no other differences in trip recovery between groups were statistically significant. **Conclusions:** Obese young adults appear to experience a higher rate of falling after tripping, and fall prevention interventions that focus on reducing the time necessary to complete the initial recovery step may help to reduce fall rates.

1. INTRODUCTION

Occupational slips, trips, and falls continue to be substantial economic and societal issues (Kemmlert et al., 2001; Layne et al., 2004; Leamon et al., 2010). These accidents account for more than 700 deaths and 200,000 injuries involving days away from work annually, with direct costs exceeding $11 billion (Liberty Mutual, 2007). Of all industries, the construction industry experiences the most fall-related mortalities (U.S. Bureau of Labor Statistics, 2009), and within this industry, falls are the leading cause of work-related deaths (CPWR, 2007). Of particular interest here are trip-induced falls, which account for 18% of injuries and 25% of workers compensation payments in construction (Lipscomb et al., 2006).

Tripping while walking momentarily impedes foot motion during the swing phase of gait, and results in forward rotation of the body/trunk about the stance foot. Averting a fall after tripping has, in essence, two requisites: arrest the body/trunk forward rotation (Grabiner et al., 2008; Grabiner et al., 1993), and prevent lower limb buckling (i.e. excessive knee/hip flexion) to maintain adequate hip height to allow subsequent stepping (Pavol et al., 2001). Trip recovery is the subsequent stepping response that an individual employs to, ideally, achieve these requisites, thereby increasing the base of support anteriorly, and providing a sufficient ground reaction force, the line of action of which contributes to decelerating the body/trunk forward rotation about the stance foot (Pavol et al., 2001). Simultaneously, a pushoff/extension response in the stance leg contributes to trunk deceleration, while also helping to prevent limb buckling and maintain adequate hip height (Pijnappels et al., 2004, 2005). As such, trunk kinematics and stepping characteristics are critical aspects of trip recovery. Two general stepping strategies have been documented during trip recovery. Trips during early swing typically elicit an elevating strategy in which the swing limb that contacts a tripping obstacle is immediately lifted over the obstacle in an attempt to complete a recovery step over the obstacle (Eng et al., 1994; Pijnappels et al., 2004). Trips during late swing typically elicit a lowering strategy, in which the swing limb that contacts a tripping obstacle is immediately lowered to the ground on the near side of the obstacle, then a step over the obstacle is attempted with the contralateral limb (Eng et al., 1994). Mid-swing perturbations can elicit both elevating and lowering strategies (Schillings et al., 2000).

Labor force demographics in the U.S. have changed dramatically over the past few decades. A reported 28% of workers are considered obese (Luckhaupt et al., 2014), which is a 20% increase over the past 10 years (Hertz, 2000). This is problematic because individuals who are obese have a higher rate of falling in general (Fjeldstad et al., 2008; Himes et al., 2012; Patino et al., 2010) and higher fall-related injuries in occupational settings (Ambrose et al., 2013; Janssen et al., 2011; Kemmlert et al., 2001; Swaen et al., 2014). A substantial proportion of these accidents are due to trip-induced falls (Amandus et al., 2012). The effect of obesity on trips and falls has received little attention despite changes in trunk mass potentially affecting trunk kinematics and stepping characteristics during trip recovery. Rosenblatt and Grabiner (2012), however, showed that the fall rate among older female adults after a laboratory-induced trip was almost twice as those who were obese (46%) compared to those who were not obese (25%), although this difference did not reach statistical significance. A better understanding of the effects of obesity on trip-induced falls can lead to strategies to reduce their occurrence.

The first goal of this study was to investigate differences in trip recovery between falls and recoveries following a laboratory-induced trip. Accomplishing this goal will provide valuable context when interpreting differences in trip recovery with obesity. We hypothesized that individuals who fell as a result of an induced trip would exhibit less favorable trip recovery measures, such as an initial recovery step that is shorter in distance or takes longer to complete, higher trunk angle or angular velocity, or lower hip height after the initial recovery step. The second goal of this study was to investigate differences in fall rate and trip recovery measures with obesity. Accomplishing this goal will help determine if trip-induced falls contribute to the higher fall rate associated with obesity, provide insight on the underlying biomechanical mechanisms contributing to trip-induced falls, and help focus future fall prevention efforts on the specific biomechanical factors contributing to falls among young workers. We hypothesized that obese participants would exhibit a higher fall rate and less favorable trip recovery measures compared to non-obese participants. We also explored differences in trip recovery measures between the elevating and lowering stepping strategies, and associations between selected trip recovery measures. Our focus was on young adults given that they comprise a large proportion of the workforce, and because the biomechanical mechanisms of trip-induced falls may differ from those among older adults (Pavol et al., 2001).

2. METHODS

Twenty-one participants completed the study including 10 non-obese adults (age 21-30 years; six females; mass 55.5-81.5 kg; stature 1.58-1.79 m; body mass index or BMI 19.4-25.7 kg/m^2), and 11 obese adults (age 20-28 years; five females; mass 80.5-115.5 kg; stature 1.58-1.83 m; BMI 29.8-42.86 kg/m^2) adults. Participants were recruited from the university population using flyers and email announcements. Exclusion criteria including any self-reported musculoskeletal, neurological, or balance disorders that would affect gait, and a change in body mass greater than 2.3 kg over the prior six months. The study was approved by the university Institutional Review Board, and all participants provided written informed consent prior to participation.

All testing was performed in a single experimental session. Participants were given a verbal overview of the protocol, and informed there was a chance they would be either tripped or slipped to elicit a fall. Participants then donned standardized athletic clothes, shoes, and a full-body safety harness to prevent impact of the knees or hands with the ground in the event of an unsuccessful trip recovery. The safety harness was attached to an overhead track using a harness spreader bar, and the length of the spreader bar was adjusted such that the tips of the fingers were approximately 5 cm from the floor when reaching for the floor, and the knees (when flexed 90 degrees) were approximately 5 cm from the floor when allowing the harness to fully support body weight. Ten practice gait trials were performed along a 10 m walkway to allow participants to become accustomed to the experimental setup and procedures. Participants were given a starting position near one end of the walkway, and asked to walk at a purposeful (slightly hurried), speed with their eyes focused straight ahead. Speed was determined using a reflective marker on the medial border of the right scapula. After completing the practice trials, participants donned wireless headphones and watched a television program on computer monitors positioned at both ends of the walkway to divert attention from tripping. A message on the screen instructed the participant when to turn around and walk to the far end of the walkway. The starting position on the walkway was varied, and trials were repeated until the dominant foot was naturally and consistently positioned approximately 4 cm prior to the concealed tripping obstacle near the middle of the walkway. During a randomly selected trial, and upon this same foot placement relative to the obstacle, the trip obstacle was manually activated by pulling a concealed rope to raise the 7-cm-high obstacle and elicit a trip. All participants were successfully tripped.

During all tripping trials, ground reaction forces were sampled at 1000 Hz from a 0.9 × 0.9 m force platform (Bertec Corporation, Columbus, OH) embedded in the walkway, and harness load was sampled at 1000 Hz using a uni-axial load cell (Cooper Instruments and Systems, Warrenton, VA). Both signals were low-pass filtered at 20 Hz (eighth-order zero-phase-shift Butterworth filter) using Matlab 2013a (The Mathworks Inc., Natick, MA). Body position was sampled at 100 Hz using a modified Helen Hayes marker set and a six-camera motion analysis system (MX-T10, Vicon Motion Systems Inc., Los Angeles, CA), and subsequently low-pass filtered at 5 Hz (8th-order, zero-phase-shift Butterworth filter).

Harness load was used to classify trip recovery outcome as: 1) a recovery when peak harness force was less than 30% of participant body weight, 2) harness-assisted when peak force was 30-50% of participant body weight, and 3) a fall when peak force exceeded 50% of body weight (Brady et al., 2000). Outcomes for one obese female and one non-obese female were deemed harness-assisted, and were removed from further analysis to avoid trials with an ambiguous outcome had the participant not been wearing the harness.

Several measures of recovery kinematics were determined. Recovery step length was defined as the anterior-posterior distance between the lateral malleoli during the stance phase of the first step over the obstacle, and the contralateral foot position after completing the first step over the obstacle. Recovery step time was defined as the time elapsed from trip onset to foot strike of the first step over the obstacle, where trip onset was identified by a peak in the acceleration of a marker on the 5^{th} metatarsal head of the perturbed limb, and touchdown of the first recovery step was identified by a vertical ground reaction force greater than 15 N. Trunk angle and angular velocity in the sagittal plane were determined at foot strike of the first step over the obstacle. Trunk angle was calculated using the angle between vertical and a line passing through markers on the right greater trochanter and the medial border of the right scapula, and considered to be 0 degrees during upright quiet standing (increasing when bowing). Trunk angular velocity was calculated using trunk angle and a finite difference method. Minimum hip height was calculated to quantify limb buckling as a possible fall mechanism (Pavol et al., 2001), and was calculated over the 300 ms after foot strike of the first step over the obstacle using the hip joint center of the lower extremity that first stepped over the obstacle. The phase of gait at which the trip was initiated was determined from the horizontal distance between markers on the right and left lateral malleoli, and expressed as a percentage of the length of the contralateral stride prior to the trip (Pavol et al., 2001).

Three types of statistical analyses were performed. First, Barnard's Exact test was used to assess differences in the fall rate between obesity groups (one-sided test), differences in stepping strategy between obesity groups (two-sided test), and differences in fall rate between stepping strategies (two sided test). Effect sizes were estimated using the phi coefficient (φ), which is a special case of the Pearson product-moment correlation since both variables are binary. Second, the non-parametric Mann-Whitney *U*-test was used to assess differences in recovery step length, recovery step time, and recovery kinematics between trip-induced falls and recoveries, between elevating and lowering strategies, and between obesity groups. This non-parametric test was used due to the relatively small number of falls that occurred (see Results). Effect sizes for these analyses were estimated using Cliff's delta (d). For random variables X and Y, the d statistic is equivalent to the probability that X is larger than Y, minus the probability that Y is larger than X (Cliff, 1993). Values of d larger in magnitude correspond to larger effects. Third, Spearman rank correlation coefficient was used to quantify the association between selected measures of recovery kinematics. Gait speed and recovery step length were normalized by individual stature prior to statistical analyses, but are reported below in units of meters/sec and meters, respectively, for clarity. Significant differences where concluded when $p \leq 0.05$, and all statistical analyses were performed using JMP 10 (SAS Institute Inc., Cary, NC).

3. RESULTS

Initial conditions of the laboratory-induced trips were consistent between the two obesity groups. Trip onset was at (median [25^{th} percentile - 75^{th} percentile]) 58.1 [54.6 - 63.6] % of stride across all trips, and did not significantly differ between falls and recoveries ($p = 0.105$) or between obesity groups ($p = 0.391$), but was significantly later for trials involving the lowering strategy (63.6 [60.6 - 66.9] % of stride) than trials involving the elevating strategy (56.8 [53.7 - 59.1] % of stride; $p = 0.030$). Gait speed was 1.60 [1.56 - 1.65] meters/sec across trials, and did not differ significantly between falls and recoveries ($p = 0.065$), stepping strategies ($p = 0.287$), and obesity groups ($p = 0.903$). Trunk angle at trip onset was 0.1 [-1.0 - 2.8] deg overall, and did not differ significantly between falls and recoveries ($p = 0.744$), stepping strategies ($p = 1.000$), and obesity groups ($p = 1.000$).

All 21 participants were successfully tripped. Of the 21 trips, three were classified as falls, two as harness-assisted, and 16 as recoveries. Trip recovery measures differed significantly between falls and recoveries (Table 1). All three falls involved the lowering strategy, while 14 out of 16 recoveries involved the elevating strategy. Recovery step length did not differ between falls and recoveries, but recovery step time was significantly longer during falls than recoveries. Trunk angle at foot strike and trunk angular velocity at foot strike were both significantly larger during falls than recoveries. Minimum hip height was significantly lower during falls than recoveries.

Fall rate was noticeably different between obesity groups in that 30% of obese participants fell while 0% non-obese participants fell, though this difference only approached significance (Table 2). One faller was characterized as a "during step" faller (peak harness force occurred prior to completing a recovery step over the obstacle), and the other two fallers were characterized as "after step" fallers (peak harness force occurred after touchdown of the first recovery step). Stepping strategy differed between obese groups in that 60% of obese participants and 89% of non-obese participants used the elevating strategy, though this difference was not significant. Though recovery step length did not differ between obesity groups, recovery step time was significantly longer among obese participants. Recovery kinematics did not differ between obesity groups, with both groups having comparable trunk angles and trunk angular velocities at foot strike, and minimum hip heights.

TABLE 1 Differences in gait and trip recovery measures between falls and recoveries. Median [1st quartile - 3rd quartile] values are reported for these two outcomes, along with *p* -values and effect sizes for differences between outcomes. Significant differences are in bold.

	Falls	Recoveries	*p*	ES
Elevating	0	14	**0.002**	0.72
Lowering	3	2		
Recovery Step Length (m)	0.85 [0.70-0.88]	0.96 [0.89-1.04]	0.162	-0.54
Recovery Step Time (s)	0.50 [0.49-0.51]	0.42 [0.35-0.46]	**0.008**	1.00
Trunk Angle FS (deg)	54.1 [52.5-56.9]	27.4 [18.6-30.9]	**0.009**	1.00
Trunk Angular Velocity FS (deg/s)	143.3 [95.6-150]	17.9[-41.2-60.4]	**0.012**	0.96
Minimum Hip Height (% stature)	44.5 [28.6-46.1]	49.6 [46.0-52.7]	**0.044**	-0.78
Speed (m/s)	1.68 [1.59-1.74]	1.60 [1.54-1.63]	0.065	0.71

Note: FS=foot strike and ES = effect size. Positive ES means falls > recoveries.

TABLE 2 Differences in gait and trip recovery measures between obese and non-obese participants. Median [1st quartile - 3rd quartile] values are reported for these two groups, along with *p* -values and effect sizes for differences between outcomes. Significant differences are in bold.

	Obese	Non-obese	*p*	ES
Falls	3	0	0.070	0.41
Recoveries	7	9		
Elevating	6	8	0.128	0.33
Lowering	4	1		
Recovery Step Length (m)	0.98 [0.87-1.10]	0.92 [0.84-0.99]	0.348	0.27
Recovery Step Time (s)	0.47 [0.42-0.49]	0.35 [0.35-0.44]	**0.011**	0.70
Trunk Angle FS (deg)	31.3 [20.8-52.9]	27.2 [16.0-30.7]	0.153	0.40
Trunk Angular Velocity FS (deg/s)	51.7 [-2.3-107.5]	20.9[-43.5-63.2]	0.391	0.24
Minimum Hip Height (% stature)	46.0 [44.4-50.8]	50.1 [47.8-53.5]	0.077	-0.51
Speed (m/s)	1.60 [1.56-1.69]	1.60 [1.54-1.62]	0.903	-0.04

Note: FS=foot strike and ES = effect size. Positive ES means obesity > non-obese.

Several measures of recovery kinematics significantly differed between elevating and lowering strategies (Table 3), and there were significant correlations between some of these measures (Table 4). Recovery step length did not differ between strategies, but recovery step time was significantly longer during the lowering compared to elevating strategy. Trunk angle at foot strike was significantly larger during lowering strategies, though trunk angular velocity at foot strike did not differ between strategies. Minimum hip height did not differ between strategies. Recovery step length and recovery step time exhibited little association, and recovery step length exhibited little association with trunk measures and hip height. Recovery step time was positively associated with trunk angle at foot strike, and negatively associated with minimum and hip height. Trunk angle at foot strike was positively associated with trunk angular velocity at foot strike, and with speed. Trunk angular velocity at foot strike was also positively associated with speed.

TABLE 3 Differences in gait and trip recovery measures between strategies. Median [1st quartile - 3rd quartile] values are reported for these two outcomes, along with *p* -values and effect sizes for differences between outcomes. Significant differences are in bold.

	Elevating	Lowering	*p*	ES
Recovery Step Length (m)	0.97 [0.87-1.06]	0.88 [0.78-0.94]	0.151	-0.46
Recovery Step Time (s)	0.40 [0.35-0.46]	0.49 [0.45-0.51]	**0.012**	0.79
Trunk Angle FS (deg)	26.2 [17.0-31.4]	52.5 [28.8-55.5]	**0.030**	0.69
Trunk Angular Velocity FS (deg/s)	22.5 [-39.1-68.0]	95.6 [-19.0-146.7]	0.247	0.37
Minimum Hip Height (% stature)	48.8 [45.7-52.9]	46.1 [36.6-51.4]	0.375	-0.29
Speed (m/s)	1.60 [1.52-1.64]	1.61 [1.59-1.71]	0.287	0.34

Note: FS=foot strike and ES = effect size. Positive ES means lowering > elevating.

TABLE 4 Correlations (Spearman's rho) between gait and trip recovery measures (significant correlations in bold).

Recovery SL (% stature)					
-0.03	**Recovery ST (s)**				
0.12	**0.58**	**Trunk Ang FS (deg)**			
<0.01	0.38	**0.79**	**Trunk AV FS (deg/s)**		
0.03	**-0.63**	-0.43	-0.44	**Min HH (% stature)**	
0.29	0.06	**0.51**	**0.55**	-0.11	**Speed (% stature/s)**

Note: SL = step length, ST = step time, AV = angular velocity, FS = foot strike, Min HH = minimum hip height.

4. DISCUSSION

The first goal of this study was to investigate differences in trip recovery between falls and recoveries following a laboratory-induced trip. The rapid execution of the trip recovery response is critical to prevent body/trunk rotation and limb buckling from progressing to a point at which averting a fall is no longer possible. We hypothesized individuals who fell as a result of an induced trip would exhibit less favorable trip recovery measures. This hypothesis was supported, in that falls were associated with a longer recovery step time, larger trunk angle and trunk angular velocity at foot strike of initial recovery step over the obstacle, and a lower minimum hip height after completing the initial recovery step. These results were generally consistent with those from older adults (Pavol et al., 2001). Based upon these results, neither of the two requisites (arresting the body/trunk forward rotation, and preventing lower limb buckling) for averting a fall were achieved when falls occurred.

Only five of 19 participants, including all three fallers, used the lowering strategy. Based upon these results, it could be surmised that the use of the lowering strategy was less effective than an elevating strategy for trips induced here and therefore contributed to the three falls. However, use of the lowering strategy seemed reasonable given that the lowering strategy is typically (although not exclusively) employed when trip onset occurs later than 50% stride (Pavol et al., 2001), and trip onset among the three fallers in the current study was at 62, 64, and 65% of stride. However, the lowering strategy has been associated with a longer recovery step time among older adults (Eng et al., 1994; Pavol et al., 2001), and the same association was found among young adults in the current study. As such, the use of an elevating strategy instead of lowering strategy may have been beneficial due to shorter recovery step times. Although not typical for late swing, two participants used an elevating strategy to successfully recover, and were tripped at 69 and 71% of stride.

The second goal of this study was to investigate differences in fall rate and trip recovery measures with obesity. We hypothesized that obese participants would exhibit a higher fall rate and less favorable trip recovery measures compared to non-obese participants. Though the difference was not statistically significant, there was a notably higher fall rate among obese participants, of 30%, compared to 0% rate among non-obese participants. In the only other study, to our knowledge, that investigated the effects of obesity on fall rates after a laboratory-induced trip, Rosenblatt and Grabiner (2012) reported respective fall rates of 46% and 25% among older obese and non-obese females. Interestingly, the difference in fall rate reported by Rosenblatt and Grabiner (2012) was also not statistically significant, perhaps due to both studies having modest sample sizes (21 in current study and 25 in Rosenblatt and Grabiner). The lower fall rate found here in both groups may be due to our focus on young adults (20-30 years) while Rosenblatt and Grabiner's participants were substantially older (~55-65 years). Nevertheless, both studies found a comparable 21-30% higher fall rate among obese participants compared to age-equivalent non-obese participants. As in the current study, Rosenblatt and Grabiner (2012) also reported a higher prevalence of a lowering strategy among obese fallers. Their results, combined with those from the current study, may be due to obese adults favoring use of the lowering strategy. Such a favoring could result from neuromuscular impairments (as discussed below) associated with obesity making the lowering strategy easier, or elevating strategy more difficult, to execute.

Rather unexpectedly, the only trip recovery measure that significantly differed with obesity was a longer recovery step time among the obese group. A post-hoc power analysis suggested that doubling the number of participants would have resulted in the trunk angle at foot strike (power = 0.279) being significantly higher among the obese group, and minimum hip height (power = 0.420) being significantly lower among the obese group. This suggests that these effects existed, but that the data from the current study was insufficient to characterize the effects as statistically different. Notwithstanding these results, several underlying neuromuscular alterations associated with obesity (Madigan et al., 2014) could have contributed to the higher fall rate and less favorable trip recovery measures with obesity. First, obesity is associated with increased trunk mass (Matrangola et al., 2008) and an anterior shift of the trunk center of mass (Corbeil et al., 2001). Both of these would increase the gravitational moment that rotates the trunk forward after tripping, and therefore increase the mechanical demands necessary to arrest trunk motion. Second, obesity is associated with reduced lower extremity strength relative to body mass (Hulens et al., 2001; Wearing et al., 2006), which can be expected to compromise the capacity to enact a quick and effective stepping response, an extension response in the non-stepping leg during the initial recovery step (Pijnappels et al., 2004), and in the stepping leg during the stance phase after completing the initial recovery step (Madigan et al., 2005), all of which can contribute to trunk motion deceleration. Third, sensory deficits could contribute to a delayed response and longer recovery step time (Wu et al., 2014).

Fall prevention strategies may help to reduce trip-induced fall rates among individuals who are obese. First, weight loss will reduce trunk mass and inertia, and therefore reduce the neuromuscular demands to avert a fall after tripping. Second, strength training may increase the neuromuscular capacity to achieve the two requisites to avert a fall after tripping. Third, task-specific balance recovery training has improved fall rates and recovery kinematics among older adults (Bieryla et al., 2007; Grabiner et al., 2014), and may provide similar benefits among obese young adults. All three fallers in the current study employed a lowering strategy, and were in the upper 85^{th} percentile of the current sample in terms of recovery step time. Task-specific training may prove more effective than strength training in promoting a faster initial recovery step, or more advantageous stepping strategy (Bieryla et al., 2007; Grabiner et al., 2014; Madigan et al., 2014). Fourth, reducing walking speed may help decrease trip-induced fall rates (Pavol et al., 1999). The three fallers in the current study were among the upper 65th percentile of the sample in terms of walking speed. Walking less quickly decreases the kinetic energy of the body at the time of the trip, which leads to a slower fall and increases the time available in which to respond and recover. Consistent with this, trunk angle and angular velocity at foot strike were positively associated with gait speed. However, it is possible that lowering gait speed has an unintended consequence of increasing the likelihood of tripping (Garman et al., 2015).

There are several limitations to this study that warrant discussion. First, anticipation effects may have existed since participants were informed that they may be tripped or slipped. However, these effects were reduced by distraction (watching a television program), performing numerous trials prior to the trip, and tripping during a randomly selected trial. Second, the marker placement used to measure trunk kinematics was sensitive to both trunk inclination angle and lumbar flexion. However, both are related to the trunk gravitational moment and momentum that must be overcome during trip recovery. Third, as with any study employing a cross-sectional design, other differences between groups besides the characteristics reported here could have contributed to the results.

In conclusion, obese young adults fell at a notably higher rate after a laboratory-induced trip compared to non-obese counterparts. This finding may help explain the higher fall rate among obese adults reported elsewhere. This higher fall rate was associated with a longer time to complete the initial recovery step, and use of a lowering strategy. As such, interventions that focus on reducing the time to complete a recovery step may reduce fall rates after tripping. Such interventions could involve task-specific training or walking less quickly.

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Obesity alters lower extremity strength similarly amongst young and older females

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ABSTRACT

BACKGROUND AND PURPOSE: Individuals who are obese exhibit a higher rate of limited mobility compared to their healthy-weight counterparts. To better understand the underlying causes of this impairment, the purpose of this study was to comprehensively assess the effect of obesity on lower extremity strength among young and older females.
METHODS: Using a cross-sectional design, 10 young (18-30 years) healthy-weight (body mass index 18-24.9 kg/m^2), 10 young obese (body mass index >30 kg/m^2), 10 older (65-80 years) healthy-weight, and 10 older obese female participants performed isokinetic maximum voluntary contractions in ankle plantar flexion (PF), ankle dorsiflexion (DF), knee extension (KE), knee flexion (KF), hip extension (HE), and hip flexion (HF).
RESULTS and DISCUSSION: Obesity increased absolute strength in DF (p=0.001), KE (p=0.003), and HF (p<0.001), and decreased strength relative to body mass in PF (p<0.001), DF (p=0.016), KE (p=0.014), KF (p<0.001), HE (p<0.001), and HF (p<0.001). The only interactive effect between obesity and age was observed for absolute HF strength (p=0.049), where young obese exhibited higher strength compared to young healthy-weight, but no effects of obesity were observed among older female participants.
CONCLUSIONS: Results suggest the effects of obesity are mostly consistent between young and older groups, and that the effects of obesity differ between joints/exertion combinations.

INTRODUCTION

Obesity is a major contemporary medical and societal concern in that over one-third of adults in the United States are obese.[1] One of the many problems associated with obesity is limited mobility,[2-4] which can include difficulty rising from a chair,[5, 6] difficulty ascending or descending from stairs,[7] lower gait speed,[8,9] and poorer balance.[10-12] This limited mobility has been related to altered lower extremity strength and increased lower extremity strength demands associated with greater body mass.[13]

Individuals who are obese exhibit altered lower extremity strength compared to healthy-weight individuals.[14-18] Lower extremity strength can be characterized in terms of absolute strength and relative strength. Absolute strength can be defined as the maximum force or net muscle moment that can be generated at a joint expressed in units of Newtons or Newton-meters. Relative strength is typically determined as absolute strength divided by body mass[14,18] or fat-free mass.[15,17] The effects of obesity on absolute and relative strength can differ substantially. For example, the knee extensors exhibit higher absolute strength,[14-17] but lower relative strength[14,17] among individuals who are obese. The increase in absolute strength is thought to be a neuromuscular adaptation to the chronic exposure to increased body weight,[14,15] since the knee extensors play a major role in supporting the body against gravity while standing and walking. Consistent with this hypothesis, absolute strength of the knee flexors, which do not play a major role in supporting the body against gravity while standing or walking, is not higher among individuals who are obese.[14,15] Most studies investigating the effects of obesity on lower extremity strength have tested only the knee musculature despite the importance of ankle and hip strength on mobility.[11,19-21] As such, it remains to be determined how obesity affects ankle and hip strength.

The prevalence of obesity among women over the age of 60 has increased from 31.5% in 2003/2004 to 38.1% in 2011/2013.[1] Since older adults exhibit limited mobility that may result from altered lower extremity strength, it is important to understand how obesity affects strength among older adults. The combined effects of obesity and aging on absolute strength are not intuitive, given the typical loss of strength associated with aging[22,23] and the gain in strength associated with obesity.[14,17,18] Among older adults, however, obesity was associated with increase in absolute knee extensor strength[24] and decrease in relative knee extensor strength.[18] Additional studies are thus needed to more fully understand the interaction of obesity and age on lower extremity strength.

The purpose of this study was to determine the effects of obesity on lower extremity strength at the hip, knee, and ankle. These data will help elucidate whether the effects of obesity differ between lower extremity joints, between exertion directions, and between young and older adults. We focused on females because of their higher prevalence of obesity,[1] higher prevalence of obesity-related mobility impairment,[4] and higher rate of falls and fall-related injuries.[25] The specific hypotheses investigated were that 1) obesity would increase absolute strength, 2) obesity would decrease relative strength, 3) the effect of obesity on absolute strength would be smaller among older participants compared to young participants, and 4) the effect of obesity on relative strength would be larger among older participants compared to young participants. The results from this study will provide a more comprehensive understanding on how obesity affects lower extremity strength, and may help in the design of strength training interventions for maintaining mobility among individuals who are obese. For example, if obesity differentially affects young or older adults, or a specific joint/exertion direction, then targeting these differential effects may maximize the benefits on strength training on mobility maintenance.

METHODS

This cross-sectional study was performed in a university biomechanics research laboratory. Forty adult females completed the study, including 10 who were young (18-30 years) and healthy-weight (body mass index, or BMI, 18-24.9 kg/m^2), 10 young and obese (BMI >30 kg/m^2), 10 older (65-80 years) and healthy-weight, and 10 older and obese (Table 1). Participants were recruited from the university and local community using web and email announcements, flyers, and newspaper advertisements. Participants were required to pass a screening to exclude individuals with self-reported neurological, cardiac, or musculoskeletal conditions that might jeopardize their safety during testing. In addition, all participants were weight stable (<2.3 kg) for the prior 6 months, and had no obvious balance problems. Participants also completed the Godin leisure-time exercise questionnaire[26] to quantify their habitual physical activity level. Participants who performed ≥ 1 hour of moderate exercise more than 3-4 times a week were excluded from the study so as to recruit participants to avoid large differences in physical activity between participants. Body fat percentage was measured using a Lange skinfold caliper (Lange skinfold caliper; Cambridge Scientific Industries, Cambridge, Massachusetts, USA) and following the manufacturer's instructions. The study was approved by the university Institutional Review Board, and written informed consent was obtained from all participants prior to participation.

Participants completed 2 experimental sessions. Knee strength was measured during the first session, and ankle and hip strength were measured during the second session. Strength was measured as the maximum net muscle moment during concentric isokinetic maximum voluntary contractions (MVCs). These MVCs were performed using a Biodex System 3 dynamometer (Biodex Medical Systems, Inc., Shirley, NY). The experimental set-up can be seen in Figure 1.

In the first session, knee strength was measured while participants were in a seated posture and were secured (using straps) with their hip flexed approximately 70 degrees. Initially, relaxed trials were performed to measure the passive elastic/gravitational moment over the entire joint range of motion. Participants were instructed to remain relaxed while the Biodex attachment moved at 5 degrees/sec through the range of motion at least 3 times. Participants then performed concentric isokinetic MVCs in knee extension (KE) and knee flexion (KF) at 75 deg/sec. The velocity was chosen to approximate values reported in previous studies. Testing was performed on the right lower extremity. The total of 4 MVCs were completed for each exertion direction. Prior to data collection, participants performed 1 practice trial for each exertion.

In the second session, ankle and hip strength were measured in this order. The testing protocol was similar to that used for the knee, with specific differences in body positions and isokinetic velocities. Concentric isokinetic MVCs in ankle plantar flexion (PF) and dorsiflexion (DF) were performed at 60 degrees/sec in a seated position, while the knee and hip were flexed 50 degrees and 80 degrees, respectively. Concentric isokinetic MVCs in hip extension (HE) and hip flexion (HF) were performed at 60 degrees/sec in a standing position, while the knee was held in a near fully extended position.[27]

Joint angle, angular velocity, and moment were sampled from the dynamometer at 200 Hz and low-pass filtered at 5 Hz (4^{th}-order Butterworth filter). The passive elastic/gravitational moment was estimated by fitting a curvilinear line (least square) to moment data from relaxed trials throughout the range of motion, and was subtracted from each MVC trial.[27] The isokinetic region was identified for each trial, defined by where the acceleration was negligible, and the peak moment in that region was determined. All the post processing was performed in Matlab (The MathWorks Inc., Natick, Massachusetts, USA). The maximum moment across the 4 trials for each joint/exertion direction was then used as the absolute strength for that joint/exertion direction. Relative strength was determined by normalizing this strength measurement to body mass.

Separate 2-way analyses of covariance were used to determine the effects of obesity (healthy-weight or obese), age (young or older), and their interaction on each strength measurement. Strength measurements included absolute and relative strength in PF, DF, KE, KF, HE, and HF. Godin score was used as a covariate. The first and second hypotheses were tested using the main effect of obesity. The main effects of age are also reported, but these were not a major focus since they have been reported in several of previous studies. The third and fourth hypotheses were tested using the age x obesity interaction. In the event of a significant obesity x age interaction simple effects testing was used to assess the effects of obesity within each age group, and the effects of age within each obesity group. Statistical analyses was performed using JMP Pro 10 (SAS Institute, Inc., Cary, NC) with a significance level of $p<0.05$. Effect sizes were reported using the partial eta squared (${\eta_p}^2$).

RESULTS

Absolute and relative strength for young and older groups are shown in Figure 2 and Figure 3. Several main effects of obesity were observed on both absolute and relative strength (see Table 2 for group means, *p*-values, and effect sizes). Absolute HF strength exhibited the only obesity x age interaction; HF absolute strength was 31% higher among obese vs. healthy-weight young females, but did not differ between obese older and healthy-weight older females (Figure 2). Absolute strength among obese females was 28% higher in DF, and 28% higher in KE compared to healthy-weight females. Relative strength among obese females was 32% lower in PF, 16% lower in DF, 17% lower in KE, 29% lower in KF, 31% lower in HE, and 21% lower in HF compared to healthy-weight females.

Several main effects of age were also observed on both absolute and relative strength (Table 2). Regarding the noted obesity x age interaction for absolute HF strength, this was 27% higher among obese young vs. obese older females, but did not differ between healthy-weight young and healthy-weight older females (Figure 2). Absolute strength among older females was 39% lower in DF, 45% lower in KE, 55% lower in KF, and 26% lower in HF compared to young females. Relative strength among older females was 31% lower in DF, 35% lower in KE, 47% lower in KF, and 19% lower in HF compared to young females.

DISCUSSION

The purpose of this study was to investigate the effects of obesity and age on ankle, knee, and hip strength. Our first hypothesis was that obesity would increase absolute strength. This hypothesis was supported for DF, KE, and HF, since all 3 were higher among obese participants. Our second hypothesis was that obesity would decrease relative strength. This hypothesis was supported for all 6 joint/flexion-extension combinations, since relative strength was lower among obese participants. Our third hypothesis was that the effect of obesity on absolute strength would be smaller among older participants. This hypothesis was supported for HF since absolute HF strength was higher among young obese compared to young healthy-weight participants, but did not differ between older obese and older health-weight participants. Our fourth hypothesis was that the effect of obesity on relative strength would be larger among older participants compared to young. This hypothesis was not supported for any joint/flexion-extension combination since no age x obesity interaction effect was observed for any of those strength measurements. More broadly, these results indicate that the effects of obesity on lower extremity strength are generally consistent between age groups (considering the small number of obesity x age interaction effects). The results also suggest that the effects of obesity differ in magnitude, in some cases substantially, between joints/flexion-extension combinations.

Strength differences at the knee measured here were generally consistent with previous reports. Absolute KE strength was found to be 28% higher among obese compared to healthy-weight females, somewhat larger than earlier differences of 12-20%.[14,15,17,18] Relative KE strength was 17% lower among obese vs. healthy-weight females here, somewhat smaller than differences of 32% in earlier studies,[14,17] and larger than a difference of 6.5% also reported.[18] No effect of obesity on absolute KF strength was observed here, which was consistent with previous literature.[14,15] However, the 29% decrease in relative KF strength among obese participants found here is lower than the 41% decrease by Capodaglio *et al.*[14] There are at least 3 factors that can contribute to such discrepancies between the current and prior studies. First, the age group in this study was different from the mentioned studies. Age ranges in the noted prior studies were 20-40 years,[14] 20 to 79 years,[18] and 18-65 years.[15] Maffiuletti *et al.*[17], indicated mean (SD) ages of 27.0 (4.1) for healthy-weight and 25.3 (5.2) years for obese participants. Second, the ranges/means of BMIs for obese participants differ between studies; 31-43 kg/m^2,[14] over 35 kg/m^2,[17] and over 26.4 kg/m^2.[18] Third, the gender of participants can cause discrepancy in the results. While Miyatake *et al.*[18], Capodaglio *et al.*[14], and Hulens *et al.*[15] had female participants, Maffiuletti *et al.*[17] only had male participants.

As noted earlier, the higher KE absolute strength among obese participants may be due to a neuromuscular training effect from chronic exposure to higher body weight.[14, 15] The KE muscle group plays a major role in supporting the body against gravity, for example while standing and walking. Interestingly, KE relative strength was lower among obese participants, both here and in prior work. Similar differences were found for DF and HF, in that absolute strength was higher, but relative strength was lower, among obese participants. In a general sense, this suggests that the gains in lower extremity absolute strength associated with obesity are smaller, proportionally, than increases in total body mass.

It is also of interest to consider differences between lower extremity joints. Peak net muscle moments at the hip during gait are fairly similar between HE and HF,[28] suggesting that neither may experience greater absolute strength gain upon chronic exposure to higher body weight. However, the higher HF absolute strength with obesity (albeit only among young participants) could be a secondary effect of the higher KE absolute strength, since the rectus femoris muscle contributes to both KE and HF. Peak net muscle moments in DF during standing and gait are typically much smaller than in PF, and not traditionally thought to play a major role in supporting the body against gravity. As such, it is unclear why DF absolute strength, and not PF absolute strength, would increase with obesity.

This study had a few limitations that should be acknowledged. First, it is possible that some participants did not generate their maximum torque despite verbal encouragement. However, participants were encouraged to do so to the extent possible. Second, the isokinetic MVCs were measured in controlled postures and velocity, therefore, the results may not be very relevant to the real life activities. Third, the PF strength measurement set-up may have increased the chance that participants used other muscle groups such as quadriceps, to generate the force although they were instructed to only use their calf muscles. Finally, the results presented here do not necessarily generalize to other populations, or other types of muscle contractions, besides those tested.

In conclusion, obesity increased absolute strength at some, but not all, lower extremity joints and flexion-extension exertion directions, and decreased relative strength at all joints and exertion directions. The effects of obesity on lower extremity strength were generally consistent between young and older females since the only obesity x age interaction effect was observed for absolute HF strength. These results also suggest that the gain in absolute strength at some lower extremity joints that is associated with obesity is smaller (proportionally) than increases in body mass.

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Table 1. Participant characteristics (median (IQR))

Group	YH	YO	OH	OO
Age (years)	20.5 (4.5)	22 (5.8)	69 (8.8)	68.5 (8.5)
Height (cm)	165 (5.7)	168.45 (7.3)	161 (10.4)	161.75 (9)
Mass (kg)	62.1 (10.3)	94.1 (9.5)	59 (8.8)	87.75 (8.5)
BMI (kg/m^2)	22.37 (3.2)	33.07 (4.3)	22.16 (5.8)	32.46 (4.2)
Body fat (%)*	23.4 (1.9)	35.3 (4.3)	33.8 (7.7)	42.6 (4.2)
Godin score	32 (19.8)	24 (30)	23.5 (20)	25.5 (20)

YH = Young and Healthy-weight, YO = Young and Obese, OH = Old and Healthy-weight, OO = Old and Obese. *determined from Lange skinfold caliper (Lange skinfold caliper; Cambridge Scientific Industries, Cambridge, Massachusetts, USA)

Table 2. *P*-values (and effect size) for the effects of obesity, age, and obesity x age on different exertions.

	Absolute Strength (N·m)			**Relative Strength (N·m/kg)**		
	Obesity	Age	Obesity × age	Obesity	Age	Obesity × age
PF	0.560 (0.011)	0.320 (0.031)	0.074 (0.096)	<0.001* (0.317)	0.990 (0)	0.196 (0.052)
DF	0.001* (0.276)	<0.001* (0.402)	0.341 (0.028)	0.016* (0.168)	<0.001* (0.334)	0.715 (0.004)
KE	0.003* (0.239)	<0.001* (0.412)	0.181 (0.055)	0.014* (0.173)	<0.001* (0.369)	0.978 (0)
KF	0.152 (0.063)	<0.001* (0.576)	0.418 (0.021)	<0.001* (0.436)	<0.001* (0.503)	0.386 (0.024)
HE	0.590 (0.009)	0.076 (0.095)	0.449 (0.018)	<0.001* (0.349)	0.218 (0.047)	0.919 (0)
HF	<0.001* (0.291)	<0.001* (0.376)	0.049* (0.116)	<0.001* (0.368)	0.003* (0.249)	0.582 (0.010)

Note: PF = plantar flexors; DF = dorsiflexors; KE = knee extensors; KF = knee flexors; HE = hip extensors; and HF = hip flexors.

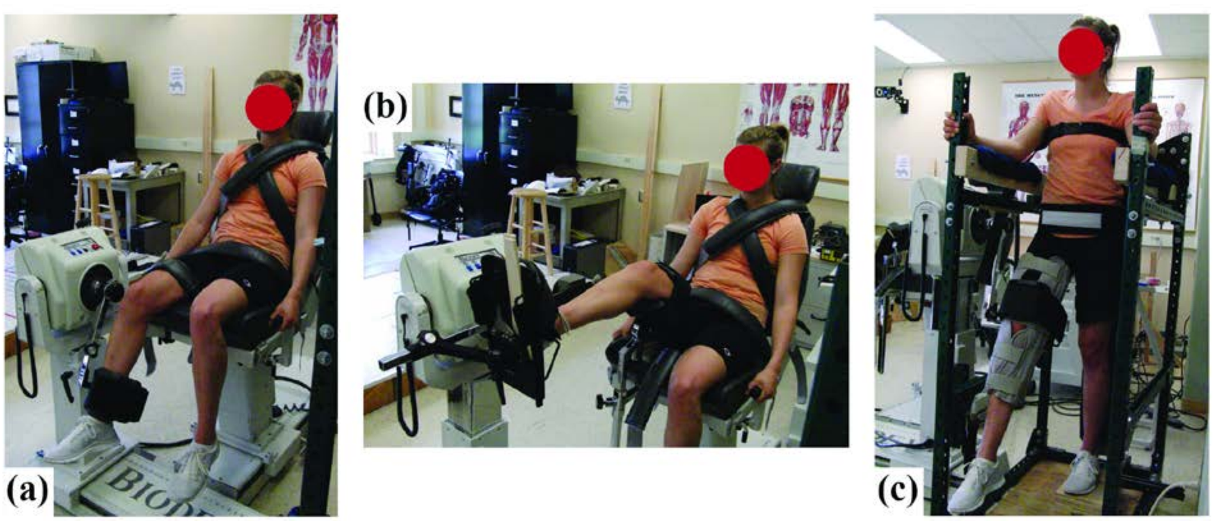

Figure 1. Experimental set-up for strength measurements at the (a) knee, (b) ankle, and (c) hip.

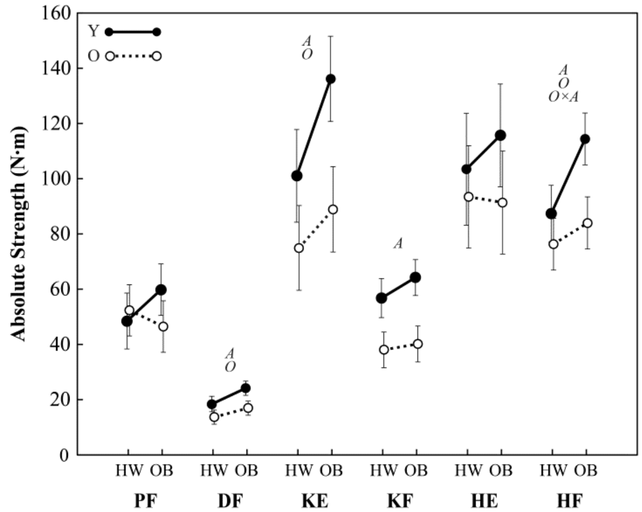

Figure 2. Absolute strength of lower extremity (error bars indicates upper and lower 95th confidence intervals). Note: A = main effect of age, O = main effect of obesity, O × A = obesity x age interaction, PF = plantar flexors, DF = dorsiflexors, KE = knee extensors, KF = knee flexors, HE = hip extensors, and HF = hip flexors.

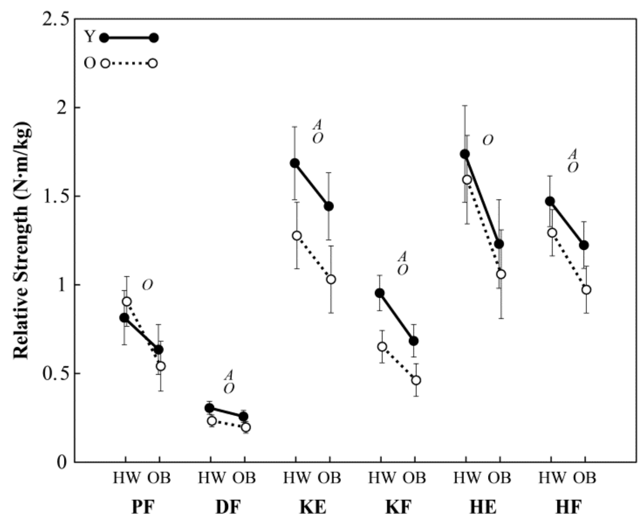

Figure 3. Relative strength of lower extremity (error bars indicates upper and lower 95th confidence intervals). Note: A = main effect of age, O = main effect of obesity, O × A = obesity x age interaction, PF = plantar flexors, DF = dorsiflexors, KE = knee extensors, KF = knee flexors, HE = hip extensors, and HF = hip flexors.

Tripping while walking elicits earlier and larger deviations in head acceleration compared to slipping

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Abstract
Slipping and tripping contribute to a large number of falls and fall related injuries among older adults. The purpose of this study was to investigate the effects of slipping and tripping on head acceleration during walking. This information can be used to better understand the potential contributions of the vestibular system to the detection and balance recovery response after slipping or tripping, and may be useful in the development of augmented sensory feedback systems. Twelve young men were exposed to an unexpected slip or trip. Head acceleration was measured and transformed to an approximate location of the vestibular system. Peak acceleration in anterior, posterior, rightward, leftward, superior, and inferior directions were compared between slipping, tripping, and walking. Peak accelerations were up to 4.68 m/s^2 higher after slipping, and up to 10.64 m/s^2 higher after tripping, compared to walking. Head acceleration first deviated from walking 100-150ms after slip onset and 0-50ms after trip onset. The temporal characteristics of head acceleration support a possible contribution of the vestibular system to detecting trip onset, but not slip onset. Head acceleration after slipping and tripping also appeared to be sufficiently large to contribute to the balance recovery response.

Introduction

Falls are of major concern among older adults due to elevated fall rates and risk for injury. An estimated 30-40% of community-dwelling older adults fall each year,[14] and falls were responsible for over 67% of non-fatal injuries and 45% of injury-related deaths among older adults in 2010.[6] Slipping and tripping accounts for an estimated 53% of falls among older adults.[19] Slipping commonly occurs while walking when the foot slips forward at heel contact, and the head/trunk subsequently falls backward. Tripping typically occurs while walking when the swing foot is obstructed, and the head/trunk subsequently falls forward. In both cases, a balance recovery response is needed to avert a fall. The more quickly the slip or trip can be detected and the recovery response initiated, the more likely a fall can be averted.

Maintaining balance is dependent upon sensory feedback from the visual, proprioceptive, and vestibular systems. Of particular interest here is the vestibular system, which detects head orientation/linear acceleration and angular acceleration using the otolith organs (utricle and saccule) and semicircular canals, respectively.[9] Sensory information from the vestibular system is utilized for upper body control across all phases of gait,[4] most likely to provide head stabilization throughout the gait cycle.[29] Head stabilization has been suggested to minimize the effect gait-related oscillations on visual and vestibular inputs, allowing a stable visual field and optimized conditions for the vestibular system.[22, 29, 30] Sensory information from the vestibular system is also utilized for lower body control during gait, and particularly during double support to assist in planning of subsequent steps.[3] Double support also allows for information from the proprioceptive system to be integrated with vestibular information to generate an internal representation of the body in space[13] and estimate if the movement of the body relative to the base of support results in the desired end position.[4]

The contribution of the vestibular system to maintaining balance varies depending upon the characteristics of perturbations to balance that are experienced. At low perturbation amplitudes (such as deviations from equilibrium during quiet standing), the vestibular system primarily contributes to balance by resolving conflicting sensory information from other sensory systems.[5] As perturbation amplitude increases, there is an increasing reliance on the vestibular system to maintain balance.[27] The location where the perturbation is applied can also influence the contributions of the vestibular system. Horak et al. showed when perturbations are applied near the head and result in early head motion, the vestibular system is primarily responsible for triggering recovery responses.[15] In contrast, when perturbations are applied more distally (e.g. the feet by a translating support surface) and result in later head motion, the proprioceptive system is primarily responsible for triggering recovery responses. Although both slips and trips involve perturbations applied to a foot, the differences in these perturbations (impulsive load applied to swing foot when tripping, and lack of friction force when slipping) could result in different time delays before alterations in head acceleration are observed. As a result, the vestibular system may contribute differently to detecting slip or trip onset.

Previous studies have shown individuals with vestibular dysfunction can be 12-times more likely to fall,[1] and the incidence of falls is greater for bilateral compared to unilateral vestibular dysfunction.[12] Such evidence supports the importance of the vestibular system in fall prevention. However, it is unclear whether the temporal and magnitude characteristics of head acceleration after slipping and tripping are amenable to the vestibular system contribution to detecting slip/trip onset as well as the subsequent balance recovery response. Knowing this would be helpful in recognizing whether individuals with vestibular dysfunction are more susceptible to slip or trip-related falls, and could contribute to the development of assistive devices providing augmented sensory feedback to replace vestibular dysfunction. Therefore, the purpose of this study was to investigate the effects of slipping and tripping on head acceleration while walking. It was hypothesized that (1) peak head acceleration after slipping and tripping would exceed those experienced while walking, and (2) deviations in head acceleration from walking would occur sooner after tripping compared to after slipping. This second hypothesis was based upon expectations that the impulsive force applied to the swing foot when tripped would affect head acceleration earlier than the lack of friction force when slipped.

Methodology

Participants

Twelve young men completed this study (mean ± standard deviation, age: 20.9 ± 2.2 years, mass: 69.9 ± 4.4 kg, height: 177.8 ± 6.3 cm). Participants were recruited from the local university population using posted advertisements and email announcements, and were required to self-report no musculoskeletal, neurological, or balance disorders that influenced gait. This study was approved by the local Institutional Review Board, and written consent was obtained from all participants prior to participation.

Experimental Procedure

Participants first performed several walking trials along a 10m level walkway covered in vinyl flooring. During these trials, participants were asked to walk naturally while looking straight ahead, and were told they would not be slipped or tripped. If mean gait speed during each trial (assessed using a motion analysis system with marker on the inferior tip of the right scapula) was not between 1.45 to 1.60 m/s, participants were asked to increase or decrease their speed and repeat the trial. Speed was constrained to prevent large variations in speed from influencing head acceleration,[24] and this speed range was selected as a purposeful (i.e. slightly hurried) walking speed for young adults.[10, 18] Participants performed 15-20 walking trials to familiarize themselves with the lab setting, feedback procedure, and gait speed, after which three trials were collected for analysis.

After walking trials were completed, participants were informed that at any point during the remainder of the session, they could be slipped or tripped as they walked down the walkway. To prevent auditory or visual cues of the slip or trip, noise protection earmuffs were worn, nature sounds were played, and the laboratory lighting was dimmed. Participants began each trial sitting on a stool at one end of the walkway with their back to the walkway. While sitting, they were given a set of letters, numbers, or symbols to memorize. The investigators then notified the participant to turn around and prepare to walk to the other end of the walkway. Once reaching the other end of the walkway, participants sat on another stool, attempted to write the memorized sequence, and then began to memorize a new sequence to remember during the next trial. This memorization task was an attempt to divert their attention away from a possible slip or trip. After repeating this for a minimum of 20 walking trials, six of the participants were unexpectedly slipped, while the other six were unexpectedly tripped, during a randomly selected trial. The right foot (preferred foot to kick a ball for all but one participant) was tripped or slipped. To elicit a slip, a foam paint roller was used to apply 50 ml of vegetable oil uniformly over a 90 x 90 cm area near the middle of the walkway while participants had their back to the walkway. Trips were induced in mid-to-late swing phase of gait using a manually actuated 7-cm-high obstacle embedded in the floor. All participants wore the same model of walking shoes in their requested size, and wore a safety harness attached to a track above the walkway to prevent impact with the floor in the event of an unsuccessful balance recovery.

Body segment positions and head acceleration were collected during all trials. Body segment positions were sampled at 200 Hz using a Vicon MX motion analysis system

(Vicon Motion Systems, Inc., Los Angeles, CA, USA), and reflective markers on the back of the head, temples, heels, lateral malleoli, and heads of fifth metatarsals. Head acceleration was sampled at 800 Hz using a lightweight (55 g) six degree-of-freedom inertial measurement unit (IMU) (Memsense, LLC., Rapid City, SD, USA) attached to the forehead using double-sided tape and wrapped with elastic cohesive bandage (Co-flex, Andover Healthcare, Inc., Salisbury, MA, USA). Marker position and acceleration data were low-passed filtered at 5 and 20 Hz, respectively (fourth-order, zero-phase-shift Butterworth filter), prior to further processing.

Estimation of Head Acceleration
Head acceleration was calculated as the linear acceleration of the right and left vestibular organs, and was expressed within a reference frame that was fixed to the head and aligned with the head's anterior-posterior (AP), medial-lateral (ML), and vertical (VER) axes when in the anatomical position. Head acceleration was calculated at the vestibular organs because it is at these locations that the body transduces accelerations into vestibular information. The location of the vestibular organs were defined to be at the intersection of an AP plane at the mastoid process, a ML plane at the infraorbitale foramen, and a VER plane at the ear canal.[16] Details of these calculations are provided in Appendix A.

To compare head acceleration between slipping and tripping with those during walking, peak acceleration in each direction (anterior, posterior, rightward, leftward, superior, and inferior) was determined over intervals from 0-50ms, 50-100ms, 100-150ms, 150-200ms, 200-250ms, and 250-300ms following perturbation onset. It was necessary to define the instant of perturbation onset for gait trials in order to compare accelerations between walking and slipping/tripping. Perturbation onset was considered to be the instant of right foot heel contact for participants who were slipped, and the instant of minimum toe clearance of the right foot for participants who were tripped.

A two-way repeated measures analysis of variance on the ranks (due to non-normal distribution of residuals) for each dependent variable (peak acceleration in the anterior, posterior, rightward, leftward, superior, and inferior directions) was used to investigate the differences between conditions (slipping/tripping or walking) and time intervals (the six intervals listed above). In the event of a significant condition by time interval interaction, contrasts were performed within each time interval to investigate differences between slipping/tripping and walking. All statistical analyses were conducted using JMP Pro 11 (Cary, North Carolina, USA) with a significance level of $p \leq 0.05$.

Results
Head acceleration during walking showed a consistent pattern both within (Figure 1) and across all participants. Head acceleration across all participants ranged from 3.14 to -4.91 m/s^2 in the AP direction (+ indicates anterior), 4.54 to -4.68 m/s^2 in the ML direction (+ indicates rightward), and 6.25 to -10.98 m/s^2 in the VER direction (+ indicates inferior). Only subtle differences in acceleration between left and right vestibular organs were measured (Figure 2) and attributed to rotations of the head and small bilateral discrepancies in the estimated location of the vestibular organs. Because

these differences were small, all subsequent analyses used the largest peak acceleration between the right and left vestibular organ.

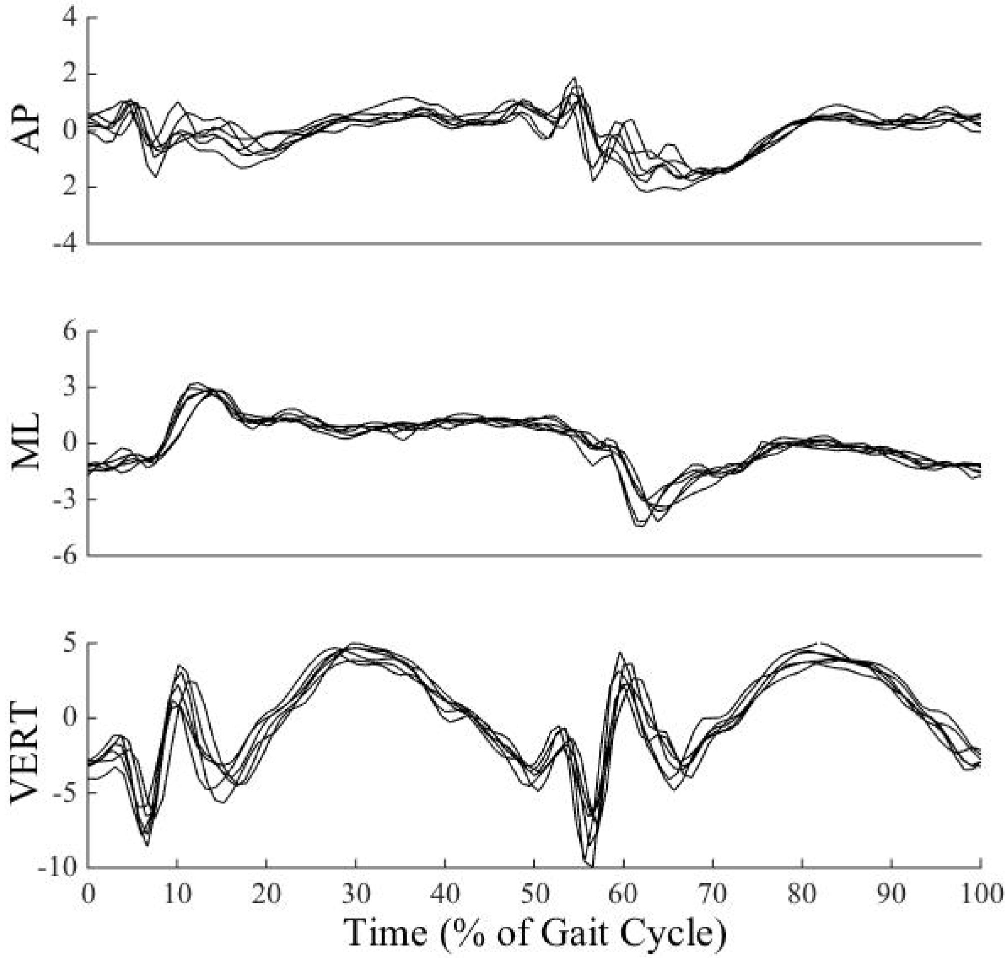

Figure 1. Acceleration (m/s^2) of the right vestibular organ during walking for a representative subject. 0% and 100% of gait cycle both represent heel contact of the left foot, while 50% of gait cycle represents the approximate time of heel contact of the right foot. Six complete gait cycles are displayed.

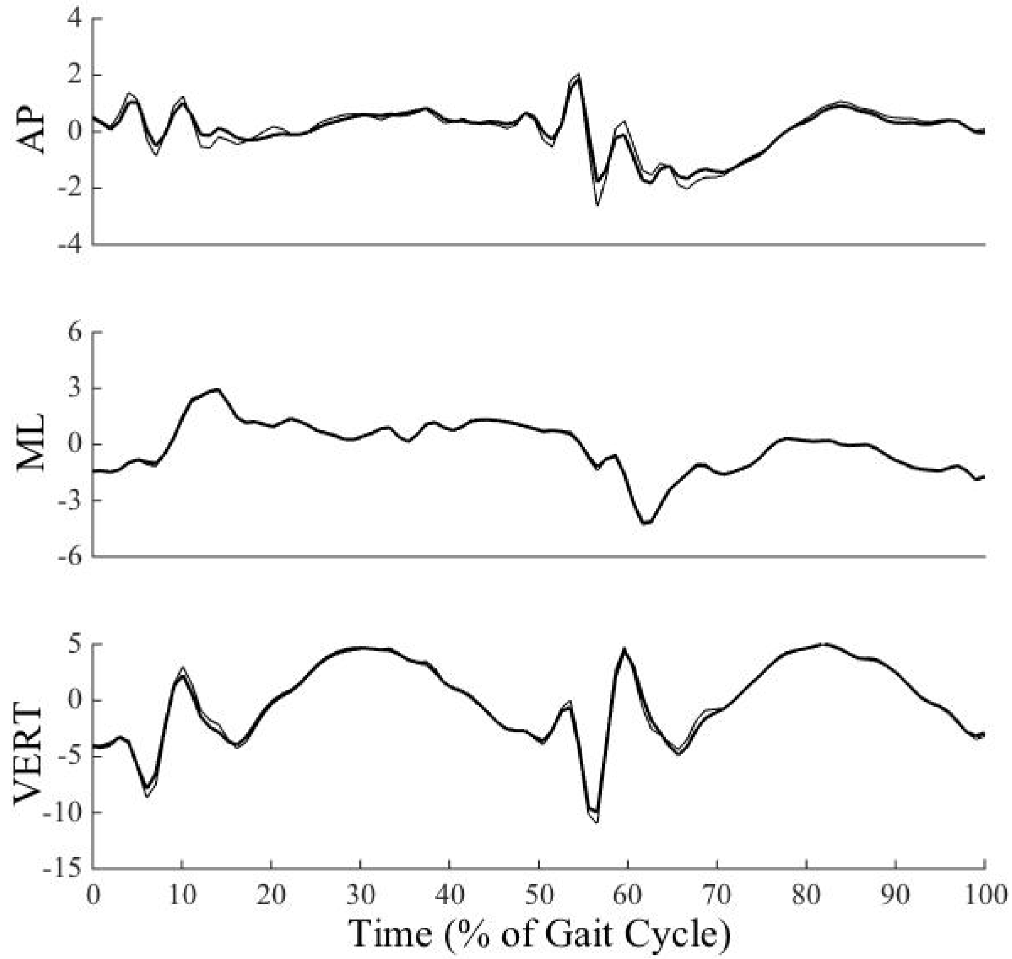

Figure 2. Accelerations (m/s^2) of the right (thick solid line) and left (thin line) vestibular organs during a selected walking stride for a representative subject. This figure illustrates the subtle differences in accelerations measured between right and left vestibular organs.

Head acceleration after slipping (Figure 3a) and tripping (Figure 3b) deviated substantially from the pattern exhibited while walking, and exhibited greater inter-participant variation compared to walking. Peak head acceleration after tripping was higher after tripping than slipping, and deviations from walking occurred more quickly after tripping than slipping.

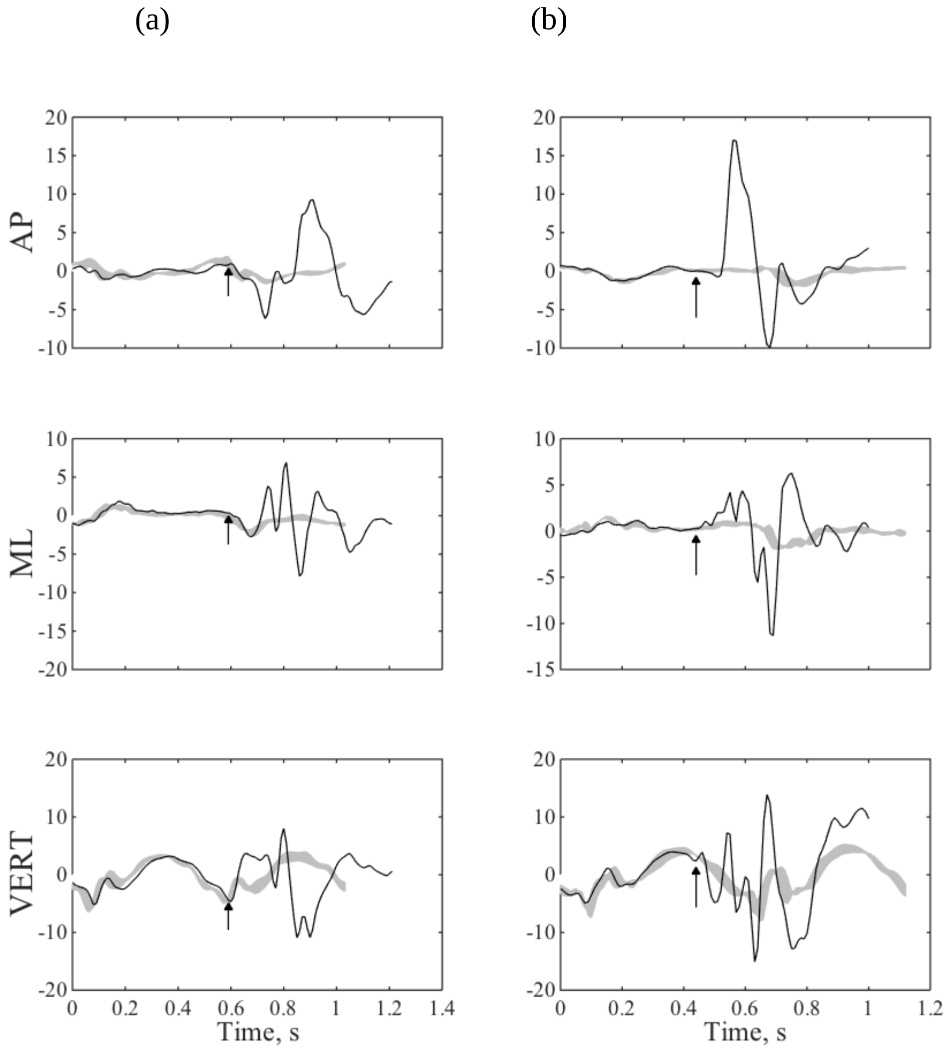

Figure 3. Head acceleration when (a) slipping (black line) and walking (gray line), and (b) tripping (black line) and walking (gray line) for a representative subject. The instant of slip and trip onset is denoted by an arrow. Head acceleration during walking is shown from left heel contact to left heel contact, and head acceleration during slipping and tripping are shown from heel contact of non-perturbed foot before perturbation (left for all participants) to subsequent heel contact of non-perturbed foot (left for all participants).

Head acceleration after slipping differed from walking in the AP, ML, and VER directions (Figure 4, actual values in Table B1). AP acceleration after slipping differed from walking in that the median peak posterior acceleration was 1.64-2.68 m/s^2 higher after slipping during the 200-250ms (p<0.001) and 250-300ms (p<0.001) time intervals. ML acceleration after slipping differed from walking in that the median peak rightward acceleration was 3.18-5.68 m/s^2 higher after slipping during the 200-250ms (p<0.001) and 250-300ms (p<0.001) time intervals. VER acceleration after slipping also differed from walking in that the median peak inferior acceleration was 2.39-4.68 m/s^2 higher after slipping during the 100-150ms (p<0.001), 150-200ms (p<0.001), and 200-250ms

(p<0.001) time intervals. Additionally, the median peak superior acceleration was 2.10 m/s^2 higher after slipping during the 250-300ms (p<0.001) time interval, but was 1.89-3.63 m/s^2 lower after slipping during the 150-200ms (p<0.001) and 200-250ms (p=0.029) time intervals. Most often, the median peak superior acceleration during the 150-200ms and 200-250ms time intervals was actually the minimum inferior acceleration due to the absence of superiorly-directed acceleration during these time intervals.

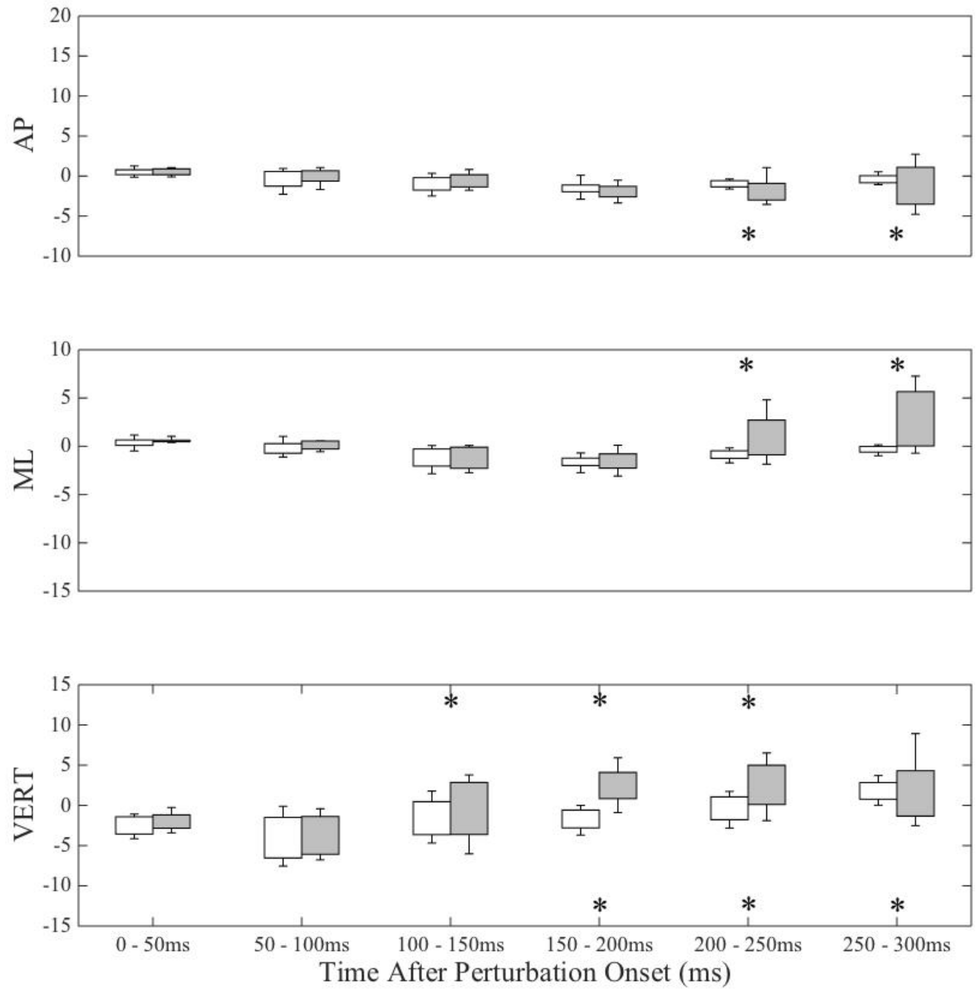

Figure 4. Peak head acceleration after slip onset (light grey) and heel contact during walking (white). The top of each column and the positive error bar indicates the median value and 75% percentile, respectively, in the positive direction (anterior, right, and inferior). The bottom of each column and the negative error bar indicates the median value and 75% percentile, respectively, in the negative direction (posterior, left, and superior). * denotes significant difference between slipping and walking within the time interval (p<0.05).

Head acceleration after tripping differed from walking in the AP, ML, and VER directions (Figure 5, actual values in Table B2). AP acceleration after tripping differed from walking in that the median peak anterior acceleration was 0.88-10.65 m/s^2 higher after tripping over all time intervals (p<0.001). In addition, the median peak posterior acceleration was 0.50-1.86 m/s^2 higher after tripping over the 0-50ms (p=0.011), 50-100ms (p<0.001), and 250-300ms (p=0.026) time intervals. ML acceleration after tripping differed from walking in that the median peak rightward acceleration was 0.74-3.20 m/s^2 higher after tripping over all time intervals (p<0.018), excluding 200-250ms (p=0.283). In addition, the median peak leftward acceleration was 2.02-5.90 m/s^2 higher after tripping over the 100-150ms (p<0.001), 150-200ms (p<0.001) and 200-250ms (p=0.004) time intervals. VER acceleration after tripping also differed from walking in that the median peak inferior acceleration was 0.70-8.85 m/s^2 higher after tripping over all time intervals (p<0.012), excluding 0-50ms (p=0.141). In addition, the median peak superior acceleration was 3.12-6.14 m/s^2 higher after tripping over all time intervals (p<0.001), excluding 200-250ms (p=0.370).

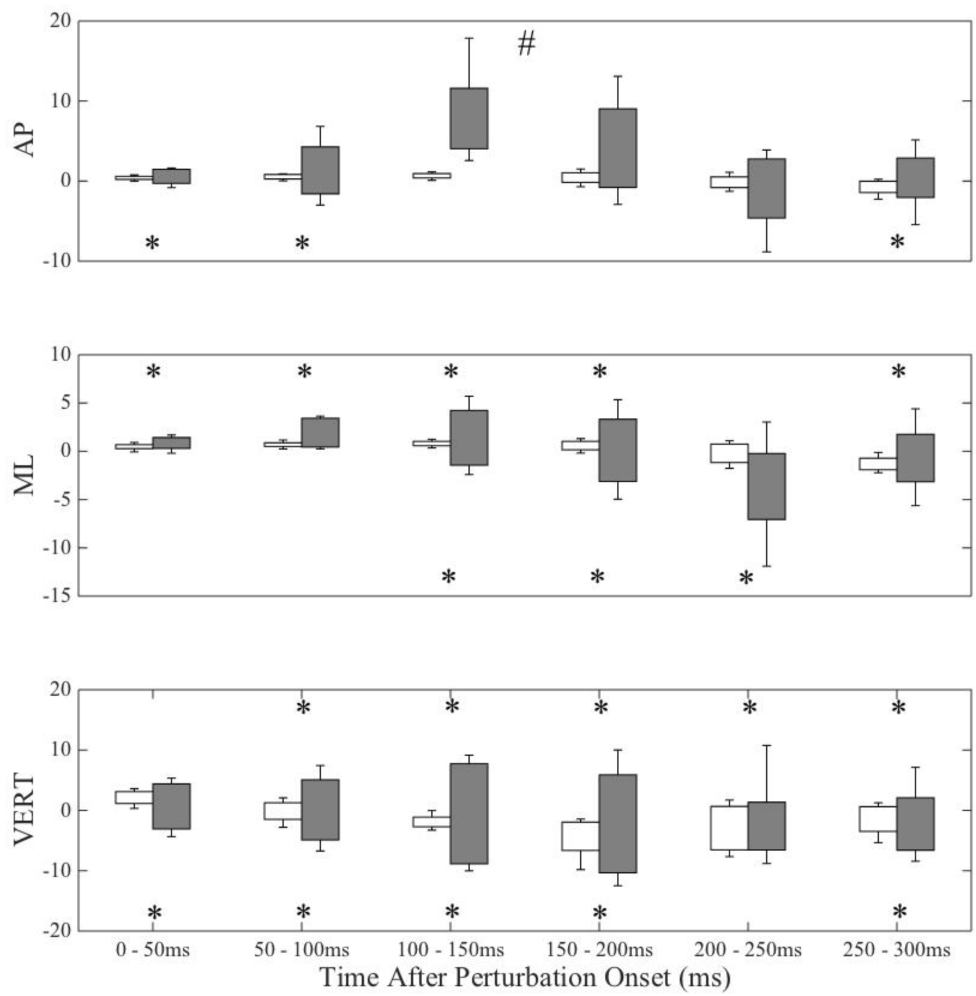

Figure 5. Peak head acceleration after trip onset (dark grey) and mid-swing during walking (white). The top of each column and positive error bar indicates the median value and 75% percentile, respectively, in the positive direction (anterior, right, and inferior). The bottom of each column and negative error bar indicates the median value and 75% percentile, respectively, in the negative direction (posterior, left, and superior). * denotes significant difference between tripping and walking within the time interval ($p<0.05$). # denotes a significant main effect of condition across all time intervals ($p<0.001$).

Discussion

The purpose of this study was to investigate the effects of slipping and tripping on head acceleration while walking. Head accelerations while walking found here were similar to previous studies.[17, 18, 22-24, 26] Root-mean-square (RMS) acceleration calculated over the entire gait cycle in the current study (compared to the range of mean values reported elsewhere) was 0.10 ± 0.04 g (0.10 to 0.11 g) in the AP direction, 0.10 ± 0.03 g (0.08 to 0.17 g) in the ML direction, and 0.29 ± 0.04 g (0.17 to 0.21 g) in the VER direction.[17, 18, 22-24, 26] The higher RMS acceleration in the VER direction of the current study was likely due to a faster walking speed (1.45-1.60 m/s) compared to prior studies (1.2-1.3 m/s). Peak head acceleration and RMS acceleration tends to increase with gait speed, and more so in the VER direction compared to AP and ML directions.[18, 24] Other small discrepancies between studies were expected due to prior studies estimating head acceleration at the posterior aspect[17] and vertex[18, 22-24] of the head.

Our first hypothesis was that peak head acceleration after slipping and tripping would exceed those experienced while walking. This hypothesis was supported because peak accelerations were up to 4.68 m/s^2 higher after slipping, and up to 10.64 m/s^2 higher after tripping, compared to walking.

Head acceleration after slipping first deviated from walking 100-150 ms after slip onset (Figure 4). This initial deviation involved the head accelerating more inferiorly compared to walking, and continued until 250-300 ms after slip onset when the head accelerated more superiorly compared to walking. Head acceleration was also more rightward (toward the foot that slipped forward) 200-300 ms after slip onset, and more posterior 200-300 ms after slip onset, compared to walking. These accelerations were consistent with head movement during a backward fall to the same side as the slipping foot, and likely result from less than expected support from the slipping leg. Previous studies have shown lower limb muscle latency times of 90 – 300 ms after the onset of a slip,[8, 21, 32] with an initial knee flexor response in the slipping leg, followed by a knee extensor response in the slipping leg.[7, 8, 20] This initial response included activity of the bicep femoris and tibialis anterior muscles, which typically have the shortest muscle latency times after slipping, ranging from 90 to 160 ms.[8, 20, 32] Because differences in head acceleration between slipping and walking reported here were not found until 100-150 ms after slip onset, these findings suggest that the vestibular system was not likely involved in detecting slip onset. However, this does not preclude a vestibular contribution later in the balance recovery response when differences from walking were found. It is also important to note that while some previous studies used similar methods to induce a slip,[7, 8] others used a sliding platform,[20, 21, 32] and differences in slipping foot kinematics between these two methods [33] could contribute to differences in muscle latency times.

Head acceleration after tripping first deviated from walking 0-50 ms after trip onset (Figure 5). Our second hypothesis was that differences in head acceleration from walking would occur sooner after tripping compared to after slipping. This hypothesis was supported because differences from walking were found 0-50 ms after trip onset, but not until 100-150 ms after slip onset. This initial deviation in head acceleration from walking involved the head accelerating more anteriorly and posteriorly, rightwardly, and superiorly compared to walking. After this 0-50 ms time interval, numerous deviations in head acceleration from walking persisted throughout the initial 300 ms after tripping that was investigated here. These accelerations were generally consistent with head movement during a forward fall to the same side as the tripped foot. However, increases in median peak head acceleration in all six directions indicated head acceleration after tripping was more complex than after slipping. Previous studies have shown muscle latency times of 55-150 ms following trip onset.[11, 28] This initial response included activity of the bicep femoris, which typically had the shortest muscle latency times, to initiate a knee flexor response in the swing leg to clear the obstacle and an hip extensor response in the stance limb to arrest forward momentum and increase the height of the body center of mass. Because differences in head acceleration between tripping and walking were found 0-50 ms after trip onset, it was plausible for the vestibular system to help detect trip onset. As after slipping, later differences from walking may have also contributed to the balance recovery response. It is also interesting to note that peak head acceleration after tripping (up to 11.59 m/s^2) exceeded those after slipping (up to 6.08 m/s^2).

In addition to deviations in head acceleration occurring early enough to contribute to onset detection and to contribute to the balance recovery response after slipping and tripping, the head acceleration differences must also be of sufficient magnitude to be detected by the vestibular system. The estimated vestibular thresholds for the detection of linear acceleration from a static initial position are 0.063 m/s^2 in AP direction, 0.057 m/s^2 in ML direction, and 0.154 m/s^2 in VER direction.[2] Median/mean head acceleration while walking reported here and elsewhere [17, 18, 22-24, 26] exceeded these thresholds at heel contact when slips commonly occur (0.18 to 0.80 m/s^2 in AP direction, 0.10 to 0.66 m/s^2 in ML direction, -3.56 to -1.42 m/s^2 in VER direction), and at mid-swing when trips commonly occur (0.21 to 0.59 m/s^2 in AP direction, 0.27 to 0.69 m/s^2 in ML direction, 1.14 to 3.11 m/s^2 in VER direction). In addition, the difference in median peak acceleration between slipping and walking, as well as between tripping and walking, at these critical times within the gait cycle also exceeded these thresholds in each direction during all time intervals (although not all differences reach statistical significance). This suggests head accelerations were of sufficient magnitude to help detect trip onset as well as the balance recovery response after slipping and tripping. However, it is unclear if the static thresholds reported above can be generalized to the dynamic situation of walking when the body does not start from a static position.

Based upon the results presented here, vestibular dysfunction is not expected to affect the detection of slip onset, but may affect the detection of trip onset. It is also possible for vestibular dysfunction to have an adverse effect on later aspects of the balance recovery response after slipping or tripping. However, sensory re-weighing is also possible to accommodate for dysfunction. The results presented here may also provide guidance in the development of sensory augmentation devices to detect losses of balance due to slipping and tripping.

Three limitations to this study warrant mention. First, we only assessed linear acceleration of the head. The vestibular system also provides feedback on head orientation and angular velocity, and these may provide additional information that contribute to the detection of slip/trip onset, and the balance recovery response. Second, anticipation-related effects may have existed when slipping or tripping. However, efforts were made to minimize any such effects using a memorization task, by requiring participants to complete at least 20 trials before being slipped/tripped, and by slipping or tripping participants unexpectedly. Third, testing was performed with young adults. Generalizing these results to other populations should be done with caution.

In conclusion, head acceleration after slipping and tripping exceeded those while walking. The temporal characteristics of head acceleration support a possible contribution of the vestibular system to detecting trip onset, but not slip onset. Head acceleration after slipping and tripping also appeared to be sufficiently large to contribute to the balance recovery response.

Appendix A.

This appendix describes the methods used to determine head acceleration at the location of the right and left vestibular organs, and within a reference frame that was aligned with the AP, ML, and VER axes of the head (Figure A1). Head acceleration was measured with the IMU worn on the forehead, and within the IMU reference frame (I-frame). The three reflective markers on the head (back of head, right temple, and left temple) were used to create a head-fixed reference frame (H-frame). These marker positions were defined within an inertially-fixed global reference frame (G-frame). The head-fixed anatomical reference frame (A-frame) was aligned with the AP, ML, and VER axes of the head.

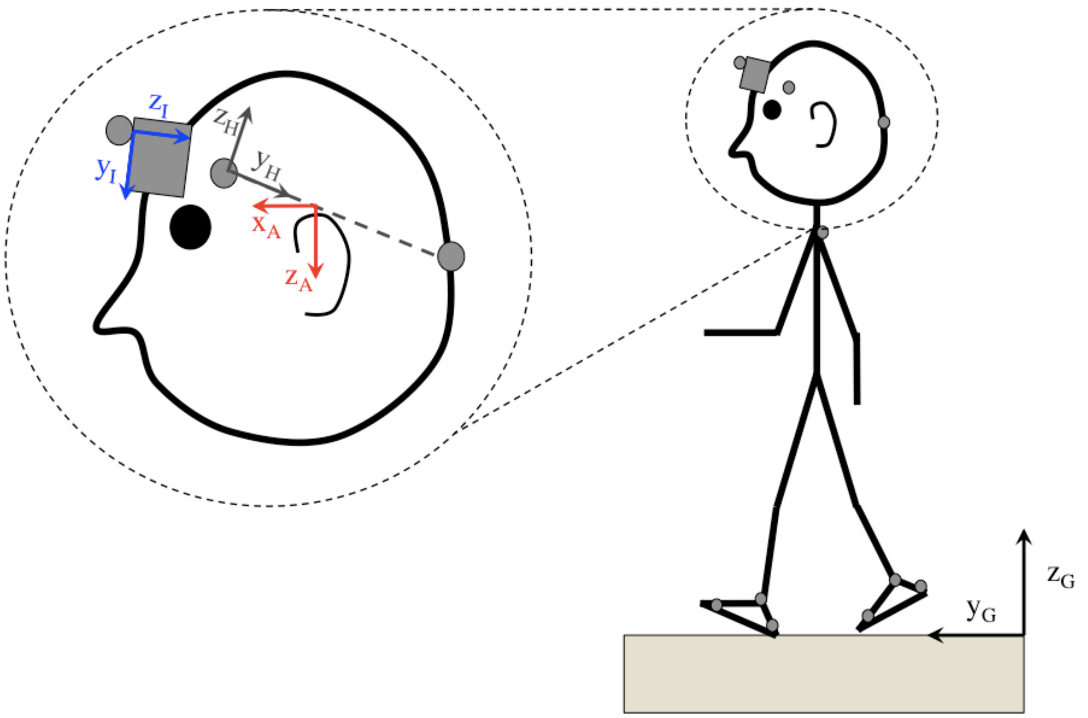

Figure A1. Diagram of the four reference frames involved in calculations including the global frame (G-frame), head-fixed reference frame (H-frame), anatomically-fixed reference frame (A-frame), and IMU-fixed reference frame (I-frame). White circles indicate the approximate position of reflective markers during testing.

Using methods similar to Rivera et al. [31], the acceleration of the vestibular organs was calculated using the five-term acceleration equation from rigid-body dynamics[25]:

$$(\vec{a}_V)_A = (\vec{a}_I)_A + \vec{\alpha} \times \vec{r}_{V/I} + \vec{\omega} \times \vec{\omega} \times \vec{r}_{V/I} + 2\vec{\omega} \times \vec{v}_{V/I} + \vec{a}_{V/I} \qquad \text{A1}$$

where $(\vec{a}_V)_A$ is the acceleration of the vestibular organ in the A-frame, $(\vec{a}_I)_A$ is the acceleration of the IMU in the A-frame, $\vec{\alpha}$ is the angular acceleration of the A-frame within the G-frame, $\vec{r}_{V/I}$ is a vector from the IMU to the vestibular organ in the A-frame, $\vec{\omega}$ is the angular velocity of the A-frame within the G-frame, $\vec{v}_{V/I}$ is the velocity of the vestibular organ in the A-frame, and $\vec{a}_{V/I}$ is the acceleration of the vestibular organs in the A-frame. It was assumed that the positions of the vestibular organs and IMU were fixed in the A-frame, and therefore $\vec{v}_{V/I}$ and $\vec{a}_{V/I}$ were set equal to zero. The methods used to calculate the terms on the right hand side of A1 are described below.

The acceleration of the IMU in the A-frame $(\vec{a}_I)_A$ was calculated by first determining the orientation of the I-frame and A-frame within the H-frame. This was done by using a two-step calibration procedure at the beginning of each session. First, a pointer with three non-collinear markers was used to define the location and orientation within the H-frame of: 1) anatomical landmarks to define the A-frame, which was aligned with the AP, ML, and VER axes of the head), 2) anatomical landmarks used to predict the location of the vestibular organs, and 3) the position of the IMU. Second, participants were then asked to hold their head stationary in three separate orientations (looking straight forward, head tilted downward to look at the floor, and then head turned to the side and slightly downward) in order to determine the orientation of the IMU (I-frame) within the H-frame. By holding the head in a static position, the acceleration collected by the IMU and transformed into the G-frame should be negligible in the AP and ML directions, and approximately 9.81 m/s^2 in the vertical direction from the acceleration due to gravity. Therefore, for the static positions tested, the acceleration of the IMU in the I-frame is:

$$\left(\vec{a}_{I,j}\right)_I = [{}_H^I R]\left(\vec{a}_{I,j}\right)_H \qquad \text{A2}$$

where $\left(\vec{a}_{I,j}\right)_I$ is the acceleration of the IMU in the I-frame of the *j*th static position, $[{}_H^I R]$ is the rotation matrix to transform linear accelerations from the H-frame to the I-frame and $\left(\vec{a}_{I,j}\right)_H$ is the acceleration of the IMU in the H-frame of the *j*th static position, which can be determined as:

$$\left(\vec{a}_{I,j}\right)_H = [{}_G^H R_j]\begin{Bmatrix} 0 \\ 0 \\ 9.81 \end{Bmatrix} \qquad \text{A3}$$

where $[{}_G^H R_j]$ is the rotation matrix to transform acceleration from the G-frame to the H-frame of the *j*th static position. Having three static positions, and determining all equations represented by the matrix equation (1), produces a system of 9 equations and 9 unknowns (i.e. vectors representing the orientation of the *x*-, *y*-, and *z*-axis of the I-frame within the H-frame). Equations containing the vector components representing the orientation of the *x*-axis of the I-frame within the H-frame were grouped into the system of equations:

$$\begin{Bmatrix} \left(a_{I,1,x}\right)_I \\ \left(a_{I,2,x}\right)_I \\ \left(a_{I,3,x}\right)_I \end{Bmatrix} = \begin{bmatrix} \left(a_{I,1,x}\right)_H & \left(a_{I,1,y}\right)_H & \left(a_{I,1,z}\right)_H \\ \left(a_{I,2,x}\right)_H & \left(a_{I,2,y}\right)_H & \left(a_{I,2,z}\right)_H \\ \left(a_{I,3,x}\right)_H & \left(a_{I,3,y}\right)_H & \left(a_{I,3,z}\right)_H \end{bmatrix} \overrightarrow{x_I} \qquad \text{A4}$$

where $\left(a_{I,j,k}\right)_m$ represents the acceleration of IMU of the *j*th static position, in the *k* direction expressed in the *m* frame, and $\overrightarrow{x_I}$ represents a vector describing the orientation of the *x*-axis of the I-frame within the H-frame. Similarly,

$$\begin{Bmatrix} \left(a_{I,1,y}\right)_I \\ \left(a_{I,2,y}\right)_I \\ \left(a_{I,3,y}\right)_I \end{Bmatrix} = \begin{bmatrix} \left(a_{I,1,x}\right)_H & \left(a_{I,1,y}\right)_H & \left(a_{I,1,z}\right)_H \\ \left(a_{I,2,x}\right)_H & \left(a_{I,2,y}\right)_H & \left(a_{I,2,z}\right)_H \\ \left(a_{I,3,x}\right)_H & \left(a_{I,3,y}\right)_H & \left(a_{I,3,z}\right)_H \end{bmatrix} \overrightarrow{y_I} \qquad \text{A5}$$

where $\overrightarrow{y_I}$ represents a vector describing the orientation of the *y*-axis of the I-frame within the H-frame. After $\overrightarrow{x_I}$ and $\overrightarrow{y_I}$ were normalized to unit vectors, the z-axis of the I-frame within the H-frame ($\overrightarrow{z_I}$) was determined as the cross product of $\overrightarrow{x_I}$ and $\overrightarrow{y_I}$. For all subsequent walking trials, acceleration was recorded by the IMU in the coordinate system aligned with the IMU itself (i.e. the I-frame) and then transformed into the A-frame by:

$$(\vec{a}_I)_A = [{}^A_H R]\,[{}^H_I R]\,(\vec{a}_I)_I \qquad \text{A6}$$

where $(\vec{a}_I)_A$ is the vector of acceleration at the IMU in the A-frame, $[{}^A_H R]$ is the rotation matrix to transform acceleration from the H-frame to the A-frame, $[{}^H_I R]$ is the rotation matrix to transform acceleration from the I-frame to the H-frame, and $(\vec{a}_I)_I$ is the vector of acceleration at the IMU in the IMU's coordinate frame. The rotation matrix $[{}^A_H R]$ was determined as:

$$ {}^A_H R = \begin{bmatrix} x_A(1) & x_A(2) & x_A(3) \\ y_A(1) & y_A(2) & y_A(3) \\ z_A(1) & z_A(2) & z_A(3) \end{bmatrix} \qquad \text{A7}$$

where x_A, y_A, and z_A are the unit vectors describing the orientation of the *x*-axis, *y*-axis, and *z*-axis, respectively, of the A-frame within the H-frame. The rotation matrix $[{}^H_I R]$ was determined as:

$$ {}^H_I R = inv\left(\begin{bmatrix} x_I(1) & x_I(2) & x_I(3) \\ y_I(1) & y_I(2) & y_I(3) \\ z_I(1) & z_I(2) & z_I(3) \end{bmatrix}\right) \qquad \text{A8}$$

where x_I, y_I, and z_I are the unit vectors describing the orientation of the *x*-axis, *y*-axis, and *z*-axis, respectively, of the I-frame within the H-frame. The angular velocity of the I-frame within the G-frame was measured with the IMU and transformed to the A-frame ($\vec{\omega}$) using rotation matrices$[{}^A_H R]$ and $[{}^H_I R]$. The angular acceleration of the A-frame in the G-frame ($\vec{\alpha}$) was calculated as the time-derivative of the $\vec{\omega}$ components.

Appendix B

Table B1. Peak head acceleration after slip onset and heel contact during walking. Values are given as: median value [25^{th} percentile, 75^{th} percentile] in m/s^2. * on slipping values denotes significant difference from walking within the time interval (p<0.05).

		0-50 ms	**50-100 ms**	**100-150 ms**	**150-200 ms**	**200-250 ms**	**250-300 ms**
Anterior	**Walk**	0.79 [0.52, 1.27]	0.57 [0.16, 0.95]	-0.19 [-1.19, 0.35]	-1.11 [-1.54, 0.11]	-0.57 [-0.90, -0.36]	0.05 [-0.31, 0.53]
	Slip	0.91 [0.41, 1.06]	0.67 [0.33, 1.06]	0.17 [-0.61, 0.83]	-1.29 [-1.51, -0.51]	-0.92 [-1.41, 1.07]	1.10 [-0.72, 2.73]
Posterior	**Walk**	0.18 [-0.23, 0.49]	-1.25 [-2.04, -0.24]	-1.75 [-2.40, -1.02]	-1.96 [-2.44, -1.03]	-1.34 [-1.85, -1.08]	-0.83 [-1.10, -0.59]
	Slip	0.17 [-0.26, 0.44]	-0.62 [-1.48, 0.44]	-1.36 [-1.85, -0.94]	-2.59 [-3.17, -1.82]	-3.00 [-5.64, -2.44]*	-3.51 [-7.47, -2.26]*
Rightward	**Walk**	0.66 [0.40, 1.17]	0.27 [0.14, 1.03]	-0.28 [-0.88, 0.08]	-1.22 [-1.63, -0.69]	-0.47 [-0.82, -0.17]	-0.02 [-0.20, 0.16]
	Slip	0.64 [0.29, 1.04]	0.55 [0.38, 0.58]	-0.10 [-1.27, 0.10]	-0.78 [-1.04, 0.12]	2.72 [0.12, 4.82]*	5.66 [1.60, 7.27]*
Leftward	**Walk**	0.10 [-0.27, 0.67]	-0.72 [-1.17, -0.33]	-2.05 [-2.64, -1.27]	-1.99 [-2.77, -1.25]	-1.25 [-1.88, -0.78]	-0.61 [-0.93, -0.24]
	Slip	0.48 [-0.21, 0.56]	-0.27 [-1.32, 0.01]	-2.29 [-2.80, -1.84]	-2.26 [-2.86, -1.43]	-0.89 [-1.17, 0.08]	0.03 [-3.05, 0.78]
Inferior	**Walk**	-1.42 [-2.29, -1.08]	-1.49 [-2.30. -0.11]	0.45 [-0.63, 1.78]	-0.59 [-1.22, 0.00]	1.05 [0.25, 1.73]	2.83 [2.34, 3.71]
	Slip	-1.18 [-2.26, -0.27]	-1.38 [-1.73, -0.42]	2.84 [1.65, 3.79]*	4.10 [2.11, 5.92]*	4.99 [4.24, 6.52]*	4.31 [3.28, 8.92]
Superior	**Walk**	-3.56 [-4.31 -2.98]	-6.54 [-7.15, -5.53]	-3.63 [-4.74, -2.59]	-2.79 [-3.71, -1.89]	-1.78 [-2.19, -0.74]	0.76 [0.13, 1.49]
	Slip	-2.83 [-4.14, -2.25]	-6.08 [-6.76, -5.38]	-3.63 [-6.41, -1.23]	0.83 [-1.44, 2.57]*	-0.11 [-1.50, 2.11]*	-1.34 [-5.27, -0.16]*

Table B2. Peak head acceleration after trip onset and mid-swing during walking. Values are given as: median value [25th percentile, 75th percentile] in m/s^2. * on tripping values denotes significant difference from walking within the time interval (p<0.05). # denotes a significant difference between tripping and walking for all time intervals (p<0.05).

		0-50 ms	50-100 ms	100-150 ms	150-200 ms	200-250 ms	250-300 ms
Anterior	**Walk**	0.59 [0.35, 0.77]	0.83 [0.50, 0.91]	0.94 [0.59, 1.16]	1.05 [0.64, 1.51]	0.53 [-0.19, 1.10]	-0.01 [-0.92, 0.24]
	Trip #	1.47 [1.00, 1.65]	4.28 [1.71, 6.83]	11.59 [10.12, 17.86]	9.03 [6.95, 13.09]	2.76 [1.48, 3.88]	2.88 [0.62, 5.14]
Posterior	**Walk**	0.21 [0.05, 0.44]	0.27 [0.05, 0.50]	0.38 [-0.00, 0.67]	-0.16 [-0.96, 0.37]	-0.80 [-1.79, -0.33]	-1.43 [-2.23, -0.59]
	Trip	-0.29 [-0.49, 0.24]*	-1.59 [-2.66, -0.19]*	4.05 [-0.11, 5.52]	-0.78 [-1.85, 1.35]	-4.62 [-8.63, -0.39]	-2.05 [-5.17, 1.34]*
Rightward	**Walk**	0.69 [0.53, 0.92]	0.88 [0.69, 1.17]	1.03 [0.80, 1.25]	1.03 [0.74, 1.32]	0.75 [0.56, 1.09]	-0.72 [-1.32, -0.12]
	Trip	1.43 [1.04, 1.69]*	3.42 [2.59, 3.65]*	4.23 [2.72, 5.71]*	3.32 [2.16, 5.34]*	-0.23 [-1.02, 3.04]	1.76 [-0.44, 4.40]*
Leftward	**Walk**	0.27 [0.09, 0.60]	0.50 [0.39, 0.76]	0.58 [0.38, 0.82]	0.16 [-0.54, 0.50]	-1.16 [-1.52, -0.56]	-1.91 [-2.45, -1.60]
	Trip	0.30 [-0.34, 0.80]	0.43 [-0.11, 0.62]	-1.44 [-4.52, -0.48]*	-3.12 [-6.21, -1.29]*	-7.06 [-9.94, -2.22]*	-3.15 [-9.76, -0.68]
Inferior	**Walk**	3.11 [1.66, 3.59]	1.26 [0.77, 2.07]	-1.12 [-1.75, -0.03]	-1.95 [-2.60, -1.44]	0.66 [-1.74, 1.72]	0.61 [-0.34, 1.27]
	Trip	4.39 [3.38, 5.35]	5.08 [3.54, 7.43]*	7.73 [4.82, 9.15]*	5.89 [1.79, 10.01]*	1.35 [0.19, 10.75]*	2.10 [0.57, 7.14]*
Superior	**Walk**	1.14 [0.60, 1.96]	-1.49 [-2.15, -0.19]	-2.71 [-3.20, -2.16]	-6.63 [-7.70, -3.45]	-6.53 [-7.35, -5.40]	-3.49 [-4.45, -1.62]
	Trip	-3.09 [-3.49, -1.84]*	-4.90 [-8.74, -3.06]*	-8.86 [-11.42, -7.70]*	-10.35 [-13.32, -8.22]*	-6.57 [-10.29, -4.34]	-6.61 [-9.98, -4.78]*

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On the number of trials to reliably determine the required coefficient of friction during gait

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1. Introduction

Slip-induced falls are a leading contributor to injuries and medical costs [1]. The most common approach to quantifying the risk of slipping while walking involves determining the peak required coefficient of friction (RCOF), either as a stand-alone measure or in relation to the available coefficient of friction at the shoe-floor interface [2, 3]. The peak value of the RCOF quantifies the minimum static coefficient of friction between the shoe and floor that is necessary to prevent the foot from slipping. Larger values of RCOF are associated with a higher risk of slipping.

As with other measures of gait, peak RCOF exhibits natural stride-to-stride variability which adversely affects its reliability [4]. Because of this, researchers commonly average (i.e. find the mean) peak RCOF values over multiple trials within the same experimental condition/session to improve its reliability. Twelve randomly-selected contemporary studies that determined RCOF during gait used as few as one, and as many as ten, trials within a given experimental condition/session. The use of multiple trials indicates an appreciation by researchers to improve reliability, but the wide range of trial numbers reflects a lack of consensus on the number of trials necessary to obtain a reliable measurement. The purpose of this study was thus to determine how the reliability of RCOF varies as function of the number of trials averaged within each experimental condition/session. The results from this work will better inform slip researchers when designing future experiments.

2. Methods

Data for this work were obtained from a prior study [5]. Subjects included 36 young male adults (mean age = 20.7±2.3 years; height = 1.80±0.08 m; mass = 79.8±11.8 kg) recruited from a university population. Subjects completed three experimental sessions, each of which was on a separate day, and all three of which were completed within one week. At the start of each session, subjects donned standardized soft-soled walking shoes to minimize variation in the frictional properties of the shoe-floor interface between subjects.

At the start of the first session, subjects were asked to complete 10 practice trials by walking at a self-selected, yet purposeful (i.e. slightly hurried), speed along a 9-meter walkway covered in vinyl flooring. These practice trials were used to determine each subject's mean self-selected speed prior to collecting data (which was required to be between 1.5 and 2 meters/second for these and subsequent slipping trials not included here). Subjects then completed 8-10 trials for data collection while maintaining the same mean speed. Trials not within ± 5% of the initial mean speed were repeated with verbal feedback given to increase or decrease speed as needed. At the start of sessions two and three, subjects were given three practice trials to adjust to the environment and re-establish their mean gait speed from the first session. Subjects then completed 8-10 additional trials during each of these sessions. Gait speed was controlled within and between sessions to minimize variations in gait speed from contributing to variations in peak RCOF [6].

During each trial, ground reaction forces were sampled at 1000 Hz using a 0.9 × 0.9 meter force platform (Bertec Corporation, Columbus, OH) positioned near the middle of the walkway, and the three-dimensional position of a single reflective marker on the right scapula was sampled at 100 Hz using a six-camera Vicon motion analysis system (T10 cameras - Vicon Motion Systems Inc., Centennial, CO). Force platform and marker position data were low-pass filtered at 7 and 5 Hz, respectively, using an eighth-order zero-phase-shift Butterworth filter. RCOF was calculated from the filtered force platform data by dividing the resultant shear force by the vertical ground reaction force, and the peak value during the initial 10-20% of stance phase of gait was used for further analysis [4]. Large absolute values of RCOF that occurred at the beginning and end of stance phase, due to small values of vertical GRF, were considered spurious and ignored. One value of peak RCOF was obtained from each trial. Gait speed for each trial was determined as the mean forward speed of the marker on the right scapula.

A three-way analysis of variance (ANOVA) was first performed to determine whether peak RCOF exhibited effects of trial, session, or their interaction (and a random subject effect). Next, the reliability of peak RCOF was quantified using the intraclass correlation coefficient (ICC) [7]. An iterative procedure was used to determine the ICC as function of the number of trials averaged (to find the mean). Initially, only the first trial from each session was used in a two-way ANOVA with session and subject (random effect) as the two factors. The $ICC_{2,1}$ was then determined from:

$$ICC_{2,1} = \frac{MS_B - MS_E}{MS_B + (k-1)MS_E + k(MS_S - MS_E)/n}$$

where MS_B = mean square between subjects; MS_E = mean square error; k = 3 sessions; MS_S = mean square between sessions; n = 36 subjects. This two-way ANOVA and ICC calculation was then repeated using the average of the first two trials from each session. This process was repeated, using progressively more trials, until the mean of all 10 trials from each session was used. The 95% confidence intervals of these ICC values were determined using prior methods [7]. All analyses were performed using R 3.2.0 and the "psych" package.

3. Results and Discussion

The overall mean (SD) of peak RCOF across all subjects and trials was 0.20 (0.02), which was consistent with values reported earlier [2, 8, 9]. As expected, peak RCOF varied for each subject, both between trials within each session, and across the three sessions (Figure 1). However, analysis of variance results indicated peak RCOF did not differ systematically between trials (p = 0.616), sessions (p = 0.108), or their interaction (p = 0.573). The ICC obtained using only the first trial from each subject was 0.58, and increased as the number of trials averaged was increased (Figure 2). Five trials were sufficient to obtain an ICC > 0.85, and appeared to be a point of diminishing return in that the increase in ICC beyond five trials was ≤ 0.01. Of note, ICC values > 0.79 has been characterized as "excellent"[10].

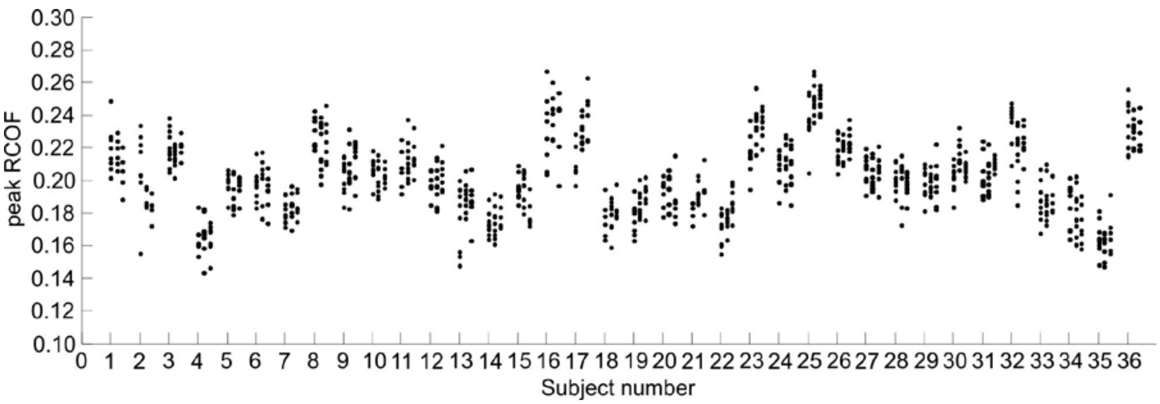

Figure 1. Peak RCOF values for all trials. The three columns of data for each subject correspond to sessions one to three when moving left to right.

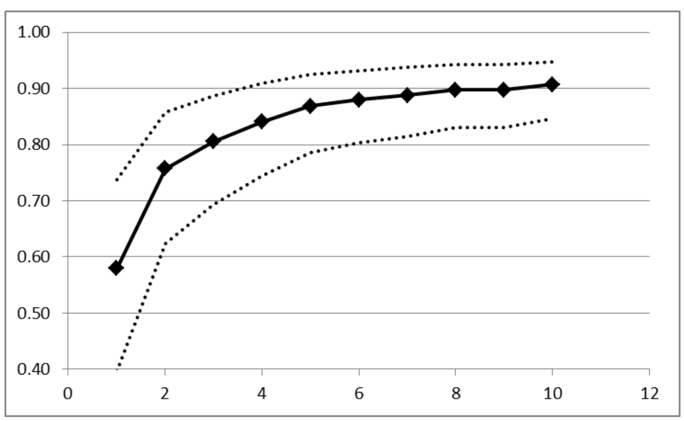

Figure 2. ICC as a function of number of trials averaged. The dotted line indicates 95% confidence interval for ICC values.

Similar to other studies [9, 11], our protocol constrained gait speed to a range of +/- 5% of each subject's self-selected speed to minimize the influence of speed on peak RCOF [12]. If gait speed is not constrained, then larger variations in gait speed than those measured here would likely reduce peak RCOF reliability, and may require averaging over more trials that those suggested here. Also, we only studied young males. Results could differ among other subject groups.

The minimum ICC value for acceptability of RCOF as a reliable measure of risk of slipping has not, to our knowledge, been established. From these data, however, it does not appear realistic to achieve ICC values above 0.90 when using a practical number of walking trials. Given this, and given the minimal improvement in ICC beyond five trials, it is suggested for slip researchers to collect and average RCOF values from 5-8 trials to obtain a reliable measure.

Conflict of interest statement
The authors have no conflict of interest to report.

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LIST OF PUBLICATIONS

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2. Allin, L.J., et al., Falls resulting from a laboratory-induced slip occur at a higher rate among individuals who are obese. Journal of Biomechanics, 2015. in review.
3. Garman, C.R., et al., Falls resulting from a laboratory-induced trip occur at a higher rate among individuals who are obese. IIE Transactions on Occupational Ergonomics and Human Factors, 2015. in review.
4. Koushyar, H., et al., Obesity alters lower extremity strength similarly amongst young and older females. Journal of Geriatric Physical Therapy, 2015. in review.
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1. Garman, Christina Maria Rossi. Understanding the effects of obesity and age on likelihood of tripping and subsequent balance recovery. PhD Dissertation. Engineering Mechanics. 2015.
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1. Allin LA, Wu X, Nussbaum MA, Madigan ML. Differences in Trailing Limb Response Between Falls and Recoveries Following a Laboratory-Induced Slip. Annual Meeting of the American College of Sports Medicine, Boston, MA, June 1-4, 2016.
2. Madigan ML, Koushyar H, Anderson DE, Nussbaum MA. Obesity increases joint moments relative to available strength during gait. Annual Meeting of the American College of Sports Medicine, Boston, MA, June 1-4, 2016.
3. Scanlon JM, Zadnik AM, Nussbaum MA, Madigan ML. Obesity does not increase likelihood of slipping while descending ramps. Annual Meeting of the American Society of Biomechanics, Columbus, OH, August 5-8, 2015.
4. Allin LJ, Wu X, Nussbaum MA, Madigan ML. Falls resulting from a laboratory-induced slip occur at a higher rate among young and older adults who are obese. Annual Meeting of the American Society of Biomechanics, Columbus, OH, August 5-8, 2015.
5. Garman CR, Franck CT, Nussbaum MA, Madigan ML. A bootstrapping method to assess the influence of gender on probability of tripping as a function of obstacle height. Annual Meeting of the American Society of Biomechanics, Columbus, OH, August 5-8, 2015.
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16. Koushyar H, Nussbaum MA, Madigan ML. The effect of obesity on hip strength through the range-of-motion among young and older females. 7th World Congress of Biomechanics, Boston, MA, July 6-11, 2014.

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19. Matrangola SL, Rossi CM, Nussbaum MA, Madigan ML. Effects of age and obesity on risk of slipping during level walking. Annual Meeting of the American Society of Biomechanics, Omaha, NE, September 4-7, 2013.
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21. Koushyar H, Bieryla KA, Madigan ML. Non-stepping balance recovery capability differs between young and older adults. Annual Meeting of the American Society of Biomechanics, Omaha, NE, September 4-7, 2013.

Cumulative Inclusion Enrollment Report

This report format should NOT be used for collecting data from study participants.

Study Title: Effects of Obesity and Age on Fall Risk - Implications for Safety Guidelines

Comments:

Racial Categories	Ethnic Categories									Total
	Not Hispanic or Latino			Hispanic or Latino			Unknown/Not Reported Ethnicity			
	Female	Male	Unknown/ Not Reported	Female	Male	Unknown/ Not Reported	Female	Male	Unknown/ Not Reported	
American Indian/ Alaska Native	0	1		1	0					2
Asian	7	16		0	0					23
Native Hawaiian or Other Pacific Islander	0	0		0	0					0
Black or African American	11	12		0	1					24
White	126	137		3	3					269
More Than One Race	1	2		0	0					3
Unknown or Not Reported	0	0		0	0					0
Total	145	168	0	4	4	0	0	0	0	321

INCLUSION OF CHILDREN

Children (defined by DHHS as individuals under the under of 21 years) were included as we recruit subjects aged 20-65 which represents the typical working age range for construction workers.

MATERIALS AVAILABLE FOR OTHER INVESTIGATORS

All experimental data are available. These may be obtained by contacting the PI, whose information is provided below. Note that the investigators wish to limit such access to individuals requesting data for research purposes only.

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