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Neurobiomechanical Mechanisms of Ballet Preventing Falls in Older Adults

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This dissertation, NEUROBIOMECHANICAL MECHANISMS OF BALLET PREVENTING FALLS IN OLDER ADULTS, by CAROLINE LAUBACHER SIMPKINS, was prepared under the direction of the candidate's Dissertation Advisory Committee. It is accepted by the committee members in partial fulfillment of the requirements for the degree, Doctor of Philosophy, in the College of Education & Human Development, Georgia State University.

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NEUROBIOMECHANICAL MECHANISMS OF BALLET PREVENTING FALLS IN OLDER ADULTS

by

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Under the Direction of Dr. Feng Yang

ABSTRACT

Falls and associated injuries are common in older adults. Dance-based interventions may be appealing options for improving balance and reducing falls in this population. Ballet emphasizes whole-body coordination and postural control, yet it is unknown if ballet practice is associated with reduced falls. The purposes of this study were to examine 1) how older ballet dancers react to an unexpected slip during standing on a treadmill compared to non-dancers, 2) if older dancers adapt to repeated treadmill standing-slips more quickly than non-dancers, and 3) if ballet practice is associated with an improvement in transfer of fall resistant skills from standing-slips to a gait-slip on the treadmill. Twenty older ballet dancers and 23 age- and sex-matched non-dancers were recruited. All participants experienced 16 standing-slips and one gait-slip on the same treadmill while full-body kinematics and bilateral leg muscle activities were recorded. Primary (dynamic gait stability) and secondary (slip outcome: fall vs non-fall, recovery stepping variables, trunk angle and angular velocity, leg muscle electromyography latency) outcomes were compared between groups. The results suggest older dancers are more stable and experience fewer falls than non-dancers in response to the first standing-slip, and dancers adapt at a faster rate to repeated standing-slips. Dancers also showed more transfer of fall resistant skills acquired from standing-slips to the gait-slip. The findings could provide insight into the neurobiomechanical mechanisms behind the association between ballet practice and reduced falls in older adults. This knowledge could also establish a basis for applying ballet as an intervention to prevent falls for older adults.

INDEX WORDS: Ballet, Dynamic gait stability, Fall prevention, Older Adults, Slip

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ABBREVIATIONS

5×STS: Five-time sit-to-stand

BF: Biceps femoris

BOS: Base of support

CI: Confidence interval

COM: Center of mass

EMG: Electromyography

FSR: Feasible stability region

LO: Liftoff

LTD: Lead foot touchdown

MG: Medial gastrocnemius

MoCA: Montreal cognitive assessment

MVIC: Maximum voluntary isometric contraction

ON: Onset

OR: Odds ratio

PD: Parkinson’s disease

PwPD: People with Parkinson’s disease

RAPA: Rapid assessment of physical activity

RF: Rectus femoris

RLO: Recovery foot liftoff

RTD: Recovery foot touchdown

TA: Tibialis anterior

TD: Touchdown

TUG: Timed-up-and-go

1 THE PROBLEM

1.1 Introduction

Falls and related injuries can result in high healthcare costs (Burns et al., 2016) and compromised quality of life (Pérez-Ros et al., 2019) in older adults. With a rapidly growing aging population (Mather et al., 2019), reducing falls is of high importance. Traditional exercise-based interventions have shown positive results in reducing fall risk in older adults (Sherrington et al., 2017), but such interventions have limitations. For example, conventional exercise programs can be physically demanding and minimally engaging, leading to low adherence rates (Farrance et al., 2016). Exergaming has also recently emerged as an encouraging alternative for eliciting balance improvements in older adults (Pluchino et al., 2012; Rendon et al., 2012; Subramaniam et al., 2021). However, as older adults become increasingly homebound and at risk for social isolation (Nicholson, 2012), interventions that include a social component become more appealing. Nearly 60% of older Americans do not meet the daily physical activity guidelines, and older adults with movement disorders are even less physically active (Keadle et al., 2016). Thus, to promote an active lifestyle, exercise modes providing benefits in an appealing and engaging manner are desired.

Dance-based interventions are an attractive option for older adults. Dance combines physical activity, musicality, socialization, and creativity in an enjoyable and safe environment (Kattenstroth et al., 2010). In addition, dance interventions exhibit high satisfaction and low attrition rates (Sharp and Hewitt, 2014; Wells and Yang, 2021a). Previous work focusing on the effects of dance-based training on fall risk factors documented that dance interventions can reduce fall risk by improving muscle strength, endurance, and balance among healthy (Blanco-Rambo et al., 2022; Harrison et al., 2024; Haussler and Earhart, 2023; Liu et al., 2021) and clinical (Dos Santos Delabary et al., 2018) older adults. Although improvement in such risk factors may lead to a decreased fall risk, fall risk factors are not the same as a fall itself (Aryee et al., 2017). This is because falls are a complex, multifactorial issue resulting from interactions among numerous intrinsic (e.g., advanced age, poor vision, diminished muscular strength, balance problems, etc.) and extrinsic (e.g., obstacles and tripping hazards, medications, slippery surfaces, etc.) risk factors (Gates et al., 2008). These previously assessed fall risk factors did not incorporate or test responses to gait perturbations most likely to initiate falls. Therefore, fall risk factors may not holistically and predictively reflect how a person responds to a perturbation (such as a slip or trip), which contributes up to 59% of older adults' falls (Simpkins and Yang, 2022b). No study has exposed older adult dancers to external perturbations and subsequent falls; thus, it remains unknown if dance-based training can indeed reduce falls. The lack of this evidence could impede the application of dance for fall prevention in older adults.

Classical ballet training emphasizes whole-body coordination, movement fluidity, and postural control while also challenging strength and flexibility (Houston and McGill, 2013). Some previous work reported better balance control in dancers than non-dancers (Golomer and Dupui, 2000), yet others argued that ballet dancers have specialized balance skills that are not necessarily transferrable to everyday situations (Hugel et al., 1999). Specific to biomechanics, Lin et al. analyzed ballet dancers' ankle biomechanics and found differences between the dominant and nondominant ankles indicating that the two ankle joints may play different roles in controlling balance (Lin et al., 2005). Additionally, another study by Lin et al. revealed worsened postural stability in injured versus non-injured dancers (Lin et al., 2011). Other work compared the center of pressure sway area, ankle moment, and muscle activity in dancers wearing new (hard) versus old (soft) pointe shoes and reported possible deleterious effects from dancing in the old (soft) shoes (Aquino et al., 2021). The contributions of the hip joint's range of motion and muscular strength in a développé à la seconde (a common movement in classical ballet involving lifting one leg very high to the side of the body) were also analyzed in young experienced ballet dancers (Metzen et al., 2023). It was concluded that the hip's strength and range of motion could predict the quality of the dancer's performance of this specific ballet step (Metzen et al., 2023). Though meaningful, these findings are somewhat limited in application to injury prevention for dancers. No study has exposed older adult ballet dancers to a well-controlled external perturbation. Therefore, it remains unclear if ballet is associated with improvements in perturbation-related fall resistance in older adults, and if so, what underlying neurobiomechanical mechanisms result in this improvement.

1.2 Research Questions

This project aimed to answer the following questions:

1. Are older ballet dancers more stable than age- and sex-matched non-dancers following the first unexpected slip perturbation during standing?
2. Do older ballet dancers adapt to repeated standing-slips more quickly than non-dancers?
3. If older ballet dancers more effectively transfer the fall resistant skills acquired during standing-slips to a gait-slip than their non-dancer counterparts?

Correspondingly, the hypotheses to be tested included:

1. Ballet dancers will experience fewer falls with better stability, a more effective recovery step, better controlled trunk movement, and shorter leg muscle electromyography (EMG) latencies after the first unexpected standing-slip compared to non-dancers.
2. Ballet dancers will show a quicker adaptation to repeated standing-slip perturbations than non-dancer controls. Specifically, dancers will exhibit greater improvement

from the first to the later standing-slips in their slip-reactions (including slip-faller rate, dynamic gait stability, trunk movement, and EMG latency) compared to the non-dancers.

3. Ballet dancers will show more transfer of the fall resistant skills learned from standing-slips to the novel gait-slip than their non-dancer counterparts. In detail, dancers will experience fewer falls with better stability, a more effective recovery step, better controlled trunk movement, and shorter leg muscle EMG latencies after the novel gait-slip compared to non-dancers.

1.3 Purpose

The overall objective of this study was to examine if and how ballet practice is associated with improvements in reactions to large-scale external slip perturbations and slip-fall risk among older adults from a neurobiomechanical perspective. The objective was threefold: 1) to examine how older ballet dancers respond to the first well-controlled and large-scale slip perturbation during standing compared to age- and sex-matched non-dancers, 2) to determine whether older ballet dancers adapt to repeated standing-slip perturbations more quickly than non-dancer counterparts, and 3) to investigate if older ballet dancers can transfer the fall resistant skills acquired from the repeated standing-slips more effectively to a novel gait-slip than older non-dancers.

1.4 Significance of the Study

This was the first study adopting well-controlled and lab-induced slips during standing and walking to examine the effects and mechanisms of ballet practice on reducing falls in healthy older adults. The association of ballet practice with both the adaptation to repeated standing-slips and the transfer of fall resistant skills to a gait slip in older adults was systematically examined. This study could fill the knowledge gap regarding the mechanisms and effects of ballet practice on reducing falls in older adults. Specifically, body kinematics and leg muscle activities were recorded during the recovery process from large-scale standing- and gait-slips. The between-group comparisons of dynamic gait stability and the slip outcome provide direct insight into the potential effects of ballet practice on preventing falls in older adults. The analyses of leg muscle EMG latency, recovery stepping, and trunk movement could offer the underlying mechanisms explaining the observations about the slip outcome from a neuromuscular and biomechanical viewpoint. A soundly understood mechanism of ballet practice reducing falls will afford a theoretical foundation for developing ballet-based fall prevention programs. As falls are a major concern in older adults, this study is practically relevant and clinically meaningful.

1.5 Delimitations and Limitations

Delimitations

This study used a minimum age of 55 years as one of the inclusion criteria. Although this age was younger than the more common standard for older adults (65 years), a minimum age of 55 has been commonly used in dance-related studies for older adults (Britten et al., 2023; Hewston et al., 2020; Marmeleira et al., 2009). This cutoff age also helped to ensure that enough participants could be recruited. Additionally, participants were only exposed to slip-based perturbations. To narrow the focus of this study, other types of perturbations (such as trips) were not explored.

Limitations

This study has limitations. First, it was difficult to recruit an equal number of male and female ballet dancers because, realistically, more females practice ballet than males. This could compromise the generalizability of our findings to both sexes. Second, to assist with recruitment, participants in this study were as young as 55 years old. While this age cutoff differs from the more standard older adult threshold of 65 years, the findings of this study can provide novel information from a neurobiomechanical perspective on how ballet practice can potentially reduce falls in older individuals. Third, this study only investigated the responses to standing- and gait-slips. It remains unclear whether the findings from the current study can be generalized to other types of perturbations such as standing-trips or gait-trips. Last, all the slips were induced on a treadmill. Therefore, it is unclear if the findings are applicable to overground perturbations. All limitations warrant more studies.

1.6 Overview of the Study

Falls are a serious public health concern in older adults (Salari et al., 2022). Although dance-based training has been employed as an alternative intervention to reduce fall risk in older adults, it remains unknown whether ballet practice can indeed reduce falls. This study adopted an observational case-control design to directly inspect if and how ballet practice may be associated with reduced fall risk in healthy older adults from a neurobiomechanical viewpoint. Twenty older ballet dancers and 23 age- and sex-matched non-dancers were enrolled in this study. Each group was exposed to repeated standing-slips followed by a novel gait-slip on a specialized treadmill. Participants' reactions to the unexpected slips were compared between dancers and non-dancers for each slip condition to test the hypotheses. This project provides insight into the neurobiomechanical mechanisms of the association between ballet practice and fall risk reduction in older adults. In addition, it could facilitate the deployment of ballet practice as an alternative intervention for preventing falls in healthy older adults and other populations with a high fall risk.

2 REVIEW OF THE LITERATURE

2.1 Falls and Consequences in Older Adults

Falls are a serious global health concern facing the older adult population (Salari et al., 2022). The risk of falling increases with age (Rubenstein, 2006), and nearly 30% of adults aged 65 and older experience at least one fall each year (Bergen et al., 2016). Falls can result in fractures and head injuries (Peel et al., 2002; Reider et al., 2024), loss of independence, reduced quality of life (Stenhagen et al., 2014), fear of falling, decreased mobility, and even death (Ambrose et al., 2013; Sherrington et al., 2020; Stevens and Sogolow, 2005). Falls also cause reduced social interactions in older adults (Yardley and Smith, 2002), leading to an increased risk of future falls and the need for long-term nursing home care (Tinetti and Williams, 1997). Over three million older adults seek hospital care for fall-related injuries each year in the United States, resulting in substantial healthcare costs (Reider et al., 2024). The estimated average cost for an inpatient fall injury was $18,658 between 2016 and 2018 (Reider et al., 2024), and medical costs for fatal and non-fatal falls in the United States in 2015 were approximately $50 billion (Florence et al., 2018). With a rapidly growing older population, the number of falls and fall-related costs are projected to increase considerably over the next 20 years (Houry et al., 2016). Strategies that assist with preventing and reducing falls in older adults are highly important and could lead to a significant reduction in fall-related healthcare spending (Florence et al., 2018).

2.2 Existing Fall Prevention Interventions

A fall is typically defined in the literature as “unintentionally coming to rest on the ground, floor, or other lower level” (Buchner et al., 1993), and the interaction of certain intrinsic (dependent on the individual) and extrinsic (dependent on the environment) risk factors can lead to an increased fall risk (Phelan and Ritchey, 2018). Some risk factors (e.g., reduced muscular strength, compromised balance, or impaired gait) can be targeted and modified using exercise-based interventions, whereas other risk factors (e.g., medication use or poor vision) require different multifactorial approaches (Sherrington et al., 2008). Over the past several decades, researchers have sought to determine how to decrease both the risk and rate of falls in older adults from an interventional perspective. Prior work investigated a wide variety of fall prevention intervention types including exercise (both alone and in combination with other intervention types), vision screening, medication management, home environment modification, fall education, and vitamin supplementation (Bischoff-Ferrari et al., 2009; Chase et al., 2012; Lord et al., 2010; Sherrington et al., 2020).

Exercise-Based Interventions

Exercise is any planned, structured, and repetitive activity that aims to improve or maintain one or more components of physical fitness (Caspersen et al., 1985). There is strong evidence that well-designed exercise-based intervention programs (particularly those that target both strength and balance) could prevent older adults from falling (Gillespie et al., 2012; Sherrington et al., 2020; Sherrington et al., 2017; Tricco et al., 2017) in a highly cost-effective manner (Davis et al., 2010; Winser et al., 2020). Exercise interventions are also effective for fall prevention in both group-based settings and on an individual basis, allowing older individuals to choose their preferred exercise scenario (Sherrington et al., 2020). According to a meta-analysis, programs involving more than three hours per week of exercise plus a component challenging balance have greater fall prevention effects for healthy older adults than programs that are less frequent and/or lack a balance component (Sherrington et al., 2017). Traditional exercise-based interventions can also reduce both the rate of falls and risk of falls per person in community-dwelling older adults (Gillespie et al., 2012). Compared with non-exercise control groups, the rate of falls in older adults who received exercise-based training can be reduced by up to 24% with balance and functional exercises, 23% with Tai Chi interventions, and 28% with interventions using multiple exercise modes (such as balance and functional exercises combined with resistance training) (Sherrington et al., 2020). However, traditional exercise-based fall prevention interventions do have limitations. For example, conventional exercise programs can be physically demanding and minimally engaging for older adults, leading to low adherence rates (Farrance et al., 2016).

Some alternative fall prevention interventions have been developed recently. For example, vibration training emerged as an encouraging modality to prevent falls for older adults (Ma et al., 2016; Wadsworth and Lark, 2020; Yang et al., 2015; Yang et al., 2023) and people with movement disorders (Alzaben and Kim, 2022; Yang, 2020; Yang and Butler, 2020; Yang et al., 2018b). Vibration training is typically performed while the trainees sit or stand on a vibration platform. The mechanical stimuli produced by the vibrating platform are transmitted to the human body and can lead to numerous physiological, biomechanical, and cognitive changes (Buehler et al., 2022; Jepsen et al., 2017; Rittweger, 2010; Yang et al., 2022b). Previous studies reported that long-term vibration training can result in improvements in a group of fall risk factors, including muscle strength, muscle power, sensation, flexibility, balance, and cognitive functions in various populations. Furthermore, one recent study also found that an 8-week vibration training program could lower the relative fall risk by 47% during lab-induced slips and lower the fall hazard by about 40% in daily living conditions among older adults (Yang et al., 2023). Although vibration training could be a promising alternative to reduce falls for older adults and other clinical populations, this training modality requires a specialized vibration platform, which could pose a potential barrier to the deployment of this type of training within the community.

Another appealing fall prevention intervention is exergaming, a combination of physical activity and video gaming (Bonetti et al., 2010). Exergaming is an encouraging and engaging alternative intervention eliciting balance improvements in older adults (Pluchino et al., 2012; Rendon et al., 2012; Subramaniam et al., 2021). Yet, exergaming is often done in solitude from the comfort and convenience of a person's own home. Older adults can increasingly become tied to their homes, putting them at a higher risk for decreased social activity and increased isolation (Nicholson, 2012). Interventions that are both engaging and include a social component therefore become more appealing. Nearly 60% of older Americans do not meet the daily physical activity guidelines, and older adults with movement disorders are even less physically active (Keadle et al., 2016). Exercise modes providing benefits in an achievable and engaging manner are desired to prompt an active lifestyle for both healthy and clinical older adults.

Multifactorial Interventions

Many fall prevention intervention programs are multifactorial in nature, utilizing an exercise-based intervention in combination with one or more other intervention types (Guirguis-Blake et al., 2018). Previous work reported that multifactorial interventions could have a significant effect on reducing falls for older adults (Chang et al., 2004; Choi and Hector, 2012). A comprehensive network meta-analysis of 54 randomized controlled trials deemed that both exercise alone and the following combinations of multifactorial interventions successfully prevented falls in older adults compared with usual care: 1) exercise, vision assessment/treatment, and environmental assessment/modification, 2) exercise and vision assessment/treatment, and 3) multifactorial assessment/treatment, calcium supplementation, vitamin D supplementation, and clinic-level quality improvement strategies (Tricco et al., 2017). Another meta-analysis of 10 studies reported that although both exercise-only and multifactorial interventions effectively reduced falls in older adults, exercise-only interventions were five times more effective than multifactorial interventions (Petridou et al., 2009). Multifactorial fall prevention interventions also have their own limitations. Specifically, they can be less cost-effective than exercise-only unifactorial interventions (Winser et al., 2020) and more labor-intensive at the individual and societal levels (Petridou et al., 2009).

2.3 Dance Therapy for Fall Prevention in Older Adults

Healthy Older Adults

Dance classes combine physical activity, musicality, socialization, and creativity in an enjoyable and safe environment, making them appealing to older adults (Kattenstroth et al., 2010). While many "virtual" dance classes emerged because of the COVID-19 pandemic, traditional dance classes are conducted in a dance studio with an in-person instructor, classmates, and occasionally even live music. The social component of dance-based interventions could be seen as an advantage over other training types such as exergaming that may lack this type of social interaction (Houston and McGill, 2013). High satisfaction and low attrition rates have also been reported for dance-based interventions (Sharp and Hewitt, 2014; Wells and Yang, 2021a).

Previous work has established that dance-based interventions can have many positive effects on older adults (Ararat-García et al., 2022; Liu et al., 2021; Rittiwong et al., 2023; Rodríguez and Paris-Garcia, 2022). According to a meta-analysis of 13 randomized controlled trials enrolling 1,029 participants, dance interventions improved physical functions including mobility and endurance in healthy older adults (Liu et al., 2021). Due to a lack of available studies, however, Liu et al. were unable to assess the effects of dance-based interventions on balance ability in healthy older adults. A systematic review of nine studies reported positive effects on physical and gait parameters for healthy older adults who participated in dance interventions of varying styles including ballroom dance, Korean dance, and line dancing (Rodríguez and Paris-Garcia, 2022). Ararat-García et al. concluded in their exploratory review that dance-based interventions had a positive effect on balance, gait, and motor functions in healthy older adults (Ararat-García et al., 2022).

Specific to the dance style of ballet, one systematic review examined the effectiveness of ballet training on health-related outcomes across a wide range of populations including children with cerebral palsy, young adult professional ballet dancers, and clinical older adult ballet dancers including people with Parkinson's disease (or PwPD), stroke, or multiple sclerosis (Letton et al., 2020). Letton et al. did not identify any studies that investigated the effects of ballet training in the healthy older adult population (Letton et al., 2020). Therefore, a large gap in the research exists regarding the effects of ballet-specific dance interventions and health-related outcomes in healthy older adults.

Dance for Older Adults with Parkinson's Disease

Parkinson's disease (PD) is a prevalent neurodegenerative disease affecting nearly one million Americans (Pringsheim et al., 2014), and balance impairment predisposes PwPD to a higher fall risk (Kerr et al., 2010; Latt et al., 2009; Nutt et al., 2011). Several previous meta-analyses investigated the effects of dance-based training on improving balance among PwPD (Carapellotti et al., 2020; de Almeida et al., 2020; Hasan et al., 2021; Lötzke et al., 2015; Meulenberg et al., 2023; Shanahan et al., 2015; Sharp and Hewitt, 2014; Tang et al., 2019). Based on various numbers of studies (ranging from 4-11) and participants (ranging from 100-393), overall positive effects (the effect size ranging from 0.45-0.96) of dance-based programs on improving balance in PwPD have been reported. Previous analyses included a multitude of dance styles such as Argentine tango (Carapellotti et al., 2020; de Almeida et al., 2020; Hasan et al., 2021; Lötzke et al., 2015; Shanahan et al., 2015; Sharp and Hewitt, 2014; Tang et al., 2019), Sardinian folk dance (Carapellotti et al., 2020; de Almeida et al., 2020; Hasan et al., 2021), Irish dance (de Almeida et al., 2020; Shanahan et al., 2015; Sharp and Hewitt, 2014; Tang et al., 2019), ballroom dance (Carapellotti et al., 2020; de Almeida et al., 2020; Shanahan et al., 2015; Sharp and Hewitt, 2014; Tang et al., 2019), Turo dance (Carapellotti et al., 2020; de Almeida et al., 2020), mixed methods dance (de Almeida et al., 2020; Hasan et al., 2021; Tang et al., 2019), and dance therapy (de Almeida et al., 2020; Tang et al., 2019).

A recent meta-analysis based on a large number of randomized controlled trials and a more rigorous calculation of effect size identified the most effective dance styles and intervention durations for improving balance in PwPD (Simpkins and Yang, 2022a). Argentine tango and Sardinian folk dance yielded the largest effect sizes. However, there was a relatively small number of available studies (n = 7 for Argentine tango; n = 1 for Sardinian folk) included in this analysis, therefore it could be premature to draw definitive conclusions about the ideal dance style for improving balance in PwPD. It was also reported that interventions (particularly of 12+ weeks) have the potential to improve balance among PwPD safely and effectively. Moreover, dance class length could play a role in determining the effects of dance-based therapies, where 90-minute classes can provide a larger effect than 60-minute classes. Such findings could inform future studies for designing effective dance-based balance training and fall prevention programs.

A few studies have investigated the efficacy of ballet-based interventions in improving trunk movement, balance, and stability in PwPD. For example, changes in trunk coordination and range of motion during gait were compared between PwPD who participated in 12 months of ballet classes and a control group of PwPD who did not take ballet classes (McGill et al., 2019a). However, no significant change across time and no significant group differences in trunk coordination and range of motion during gait were observed. This research group also compared gait variability and balance confidence in these same participants and found no significant effects of weekly ballet classes on these variables in PwPD (McGill et al., 2019b). These findings differ from other recent work that indicated dance-based interventions elicited improvements in balance in PwPD (Duncan and Earhart, 2012; Hackney and Earhart, 2009; Solla et al., 2019). This difference could be due to the frequency of the dance training, as most of the previous studies reporting balance improvements in PwPD had participants in dance classes two to three days per week (Duncan and Earhart, 2012; Hackney and Earhart, 2009; Solla et al., 2019), whereas McGill et al. had participants attend ballet classes once a week. Additionally, the participants from the studies by McGill et al. (McGill et al., 2019a, b) were not randomized into the ballet training group. These individuals were already taking regular weekly ballet classes that began before the onset of the study and data collection period. This could have impacted the results; however, the participant baseline data did not indicate any group differences in the outcome measures pre-intervention. More high-quality and randomized studies are needed to better explain the relationship between ballet training and PwPD.

Dance and Other Clinical Populations

Some work has investigated dance-based interventions for improving physical functioning in other clinical older adult populations besides PwPD. For example, a non-randomized study assessing balance, gait, and ataxia in eight females with multiple sclerosis reported a reduction in ataxia and improvements in both static and dynamic balance for participants who took part in a 16-week ballet-based program (Scheidler et al., 2018). Additionally, a review article investigated the use of dance interventions for improving health-related outcomes in post-stroke individuals (Kipnis et al., 2022). This review was conducted on 18 studies and reported that dance may facilitate improvements in balance and fall risk in this population. However, further research involving control groups is needed to validate these claims.

Overview of Classical Ballet Training

Ballet is one style of dance training that can improve health-related outcomes and fall risk factors such as muscular strength and balance in various populations (Letton et al., 2020). Classical ballet technique originated during the Italian Renaissance in the 15th century and was later further developed into its more current form in France and Russia. A traditional ballet class typically lasts between 60 to 90 minutes and consists of training at the barre followed by training "in the center" and across the floor. A ballet barre is a wooden or metal horizontal rod that is either mounted to the wall or is free-standing (supported by vertical rods) at a height approximately between the dancer's hip and shoulder heights. The barre provides the dancer with an additional source of support during the dance training process. Training at the barre focuses on learning to balance in various positions (both standing on two legs and on one leg), transferring the body weight from one leg to another, coordinating the upper body (head, arms, trunk) with the lower body (hips, knees, ankles, feet), and developing strength and control while moving in all directions (forwards, backwards, side-to-side, and rotational).

Beginner-level ballet students perform most of their barre training while facing the barre and holding on with two hands (Kostrovitskaya and Pisarev, 1978; Warren and Cook, 1989). As a ballet dancer becomes more experienced and skilled (intermediate or advanced level), the barre work progresses to include training while facing sideways to the barre and holding on with one hand, as well as executing more challenging steps and positions (Kostrovitskaya and Pisarev, 1978; Warren and Cook, 1989). After approximately 45 to 60 minutes of barre training, the remainder of the ballet class takes place "in the center," which is the large open area of the ballet studio space. This portion of the ballet class includes practicing steps and balancing in positions now without the support of a barre, executing turns and jumps, and practicing combinations of steps that involve traveling back and forth across the studio space in all directions. Additionally, ballet classes are accompanied by music that can provide auditory cues for movement initiation and rhythm. The music also allows the students to synchronize their movements with the instructor and one another.

2.4 Fall Risk Factors vs. Falls

Previous work documented that dance-based interventions can reduce fall risk by improving muscle strength, endurance, and balance among healthy (Blanco-Rambo et al., 2022; Liu et al., 2021) and clinical (Dos Santos Delabary et al., 2018) older adults. Although improvement in such risk factors could reduce the risk of falling, fall risk factors are not the same as a fall itself (Aryee et al., 2017) given that falls are a complex, multifactorial occurrence resulting from interactions among numerous intrinsic and extrinsic risk factors (Gates et al., 2008). These previously assessed fall risk factors did not incorporate or test responses to gait perturbations that are likely to initiate most falls. Therefore, fall risk factors may not holistically and predictively reflect how a person responds to a perturbation (such as a slip or trip), which contributes up to 59% of older adults' total falls (Simpkins and Yang, 2022b). No study has yet exposed older ballet dancers to external perturbations and subsequent falls; thus, it remains unknown if ballet-based training can indeed reduce falls.

Lab-induced external perturbations provide a standard platform to examine the mechanisms of how or why dance training may reduce balance loss and falls. One previous study exposed young ballet dancers to an "in-place" perturbation where the standing surface was tilted in the sagittal plane and measured EMG of the bilateral tibialis anterior and medial gastrocnemius

(Simmons, 2005). The authors reported that dancers displayed a faster long-latency neuromuscular response than non-dancer subjects when the balance was disturbed in this manner (Simmons, 2005). Although this finding was meaningful in providing preliminary insight into the balance control mechanism adopted by ballet dancers during a small "in-place" perturbation, it did not involve an investigation of the body's reactions to a perturbation requiring a recovery step. Lab-induced perturbations have also been widely used to investigate falls and fall risk factors in various populations in a safe and controlled environment (Pai et al., 2014; Pai and Bhatt, 2007; Yang et al., 2013; Yang et al., 2017). In the current study, lab-induced standardized slips (both while standing and during gait) on a treadmill were utilized to trigger backward recovery stepping responses, thus expanding the understanding of ballet dancers' balance control from static to dynamic conditions. Therefore, the current study's design is more relevant to fall prevention given that falling is a fast and dynamic process often requiring rapid recovery stepping to avoid the fall (Qiao and Yang, 2020).

2.5 Reaction, Adaptation, and Transfer (or Generalization)

Perturbations that are novel, unexpected, and large in magnitude are most likely to lead to a fall (Pavol et al., 2004a). In this scenario, a person cannot make anticipatory (or proactive) adjustments because perturbation is a new experience, and task-specific motor learning from repeated perturbations has not yet taken place. When a person experiences an external perturbation that leads to balance loss, the typical reaction to avoid a fall is to take one or more recovery steps and/or grab onto any nearby available structures (Pai and Bhatt, 2007). By analyzing an individual's reaction to this first perturbation exposure, insight can be obtained regarding neuro-biomechanical factors that may contribute to a successful (or unsuccessful) balance recovery. Furthermore, it may be particularly meaningful to compare the novel and unexpected perturbation reactions of different populations (e.g., young vs old, active vs inactive, trained vs untrained), given this first response offers somewhat of a "baseline" assessment that could indicate the favorable (or unfavorable) fall resistance qualities of that population.

Perturbation training has been widely used in healthy adults and populations with movement disorders to deliver repeated slip perturbations that mimic "real-life" falling scenarios, providing the opportunity for adaptive motor learning (Brown et al., 2023; Coelho et al., 2022; Pai and Bhatt, 2007). This type of training involves a trial-and-error process that prompts the central nervous system to make adaptive responses in motor behavior to improve the slipping response (Pai and Bhatt, 2007). Previous work has established that humans are able to adapt to unexpected or sudden changes in their environment during various tasks such as standing or walking (Marigold and Patla, 2002; Owings et al., 2001). Additionally, it has been reported that humans can maintain this capability for adaptation even into older adulthood (Pavol et al., 2002).

Adaptation to external perturbations can be achieved by improvements in proactive (feedforward) and/or reactive (feedback) control. Proactive adjustments occur before (in anticipation) of the perturbation onset and function to counteract the expected destabilizing effect of the perturbation (Yang et al., 2016). In contrast, reactive adjustments occur after (in response) to the perturbation. Proactive and reactive adaptations both play an important role in fall prevention. Reactive adaptations may reduce the chance that a balance loss will lead to a fall, whereas proactive adaptations can potentially eliminate a balance loss (Pavol et al., 2004b). In some cases, substantial improvements in proactive control can diminish the need for much (or any) reactive control (Liu et al., 2017; Pavol and Pai, 2002). However, if proactive strategies are not sufficient to overcome a perturbation, reactive corrections may also be required. Previous work established that perturbation training in healthy older adults can lead to reactive adaptations which result in improved stability and better recovery stepping, and therefore a decreased fall risk (Chien and Hsu, 2018; Grabiner et al., 2012; Pai et al., 2010; Pai et al., 2014).

The central nervous system also plays an important role in inter-task generalization, the human body's ability to take motor adaptations learned from one scenario and transfer or generalize them to a different scenario (Lam and Dietz, 2004; Seidler, 2004). In one prior study, healthy young adults successfully transferred motor adaptation gained from repeated sit-to-stand slips to gait-slips as evidenced by a lower incidence of falls during a gait-slip following perturbation training (Wang et al., 2011). Another study documented the transfer of fall resistant skills acquired during standing-slip training on a treadmill to the gait-slip test over the ground (Yang et al., 2018c). Generalization from perturbation training has also been observed across different walking surfaces. For example, repeated gait-slip perturbation training conducted on a treadmill was successfully transferred to overground gait-slips in both young adult (Yang et al., 2013) and older adult (Lee et al., 2016) participants. Given that most falls occur during locomotion (Berg et al., 1997; Simpkins and Yang, 2022b), it is meaningful to compare different populations' abilities for inter-task generalization between tasks such as standing and walking to develop effective fall prevention interventions.

Classical ballet training requires frequent multidirectional stepping while maintaining an upright posture (Clippinger, 2007), and effective recovery step execution and trunk movement control are essential to balance restoration after experiencing a slip (Yang et al., 2012). It remains unknown if ballet training in healthy older adults can improve 1) the reactive response to a novel and unexpected standing slip, 2) the adaptation response to multiple standing-slip perturbations, and/or 3) the generalization of anti-fall skills learned from standing-slips to a gait-slip relative to non-dancers of the same age and sex. Information gained from the comparison of these three types of motor learning aspects (i.e., reaction, adaption, and generalization) between older adult ballet dancers and non-dancers could provide more insight into the neurobiomechanical mechanism(s) of how and why ballet training may affect resilience to balance loss and fall risk reduction in this population.

2.6 Feasibility Stability Region Framework

Bipedal human motion is intrinsically unstable due to the body's multi-segmental inverted pendulum structure with a high center of mass (COM) resting on a small base of support (BOS). Conventionally, stability limits are defined solely by the position of one's COM relative to their BOS. For a person who is standing (a typical static task), balance can be preserved if the COM projection is within the BOS (Figure 2.1a). However, the static stability limits do not apply to dynamic tasks. For example, from toe-off to the early stance phase during walking, the COM is always behind or outside the BOS, which is formed by the contact area between the leading foot and the ground. According to the static stability limit, a person should encounter a backward balance loss in this scenario as the COM is behind the BOS. However, such a backward balance loss would not occur during normal gait.

The concept of static stability has been extended to dynamic conditions of the human body via various metrics. For example, maximum Floquet multipliers (Bruijn et al., 2013; Dingwell and Kang, 2007; Siragy and Nantel, 2018) have assessed the body's orbital dynamic stability and Lyapunov exponents (Bruijn et al., 2013; Dingwell and Cusumano, 2000; Siragy and Nantel, 2018) have evaluated local dynamic stability. Additionally, the margin of stability considers both the COM position and velocity related to the BOS to gauge the body's stability using a simple 2-link human model (Hof et al., 2005). Another metric that has been employed broadly is dynamic gait stability as defined by the Feasible Stability Region (FSR) theory (Pai and Patton, 1997; Yang et al., 2007). The FSR theory considers the COM position and velocity related to the BOS simultaneously, and it was analytically derived by computer simulation using a 7-link asymmetrical human model (Yang et al., 2007, 2008a) and experimentally verified by large amounts of human data (Yang et al., 2008a; Yang et al., 2008b).

According to the FSR, a person's control of COM velocity is as vital for dynamic gait stability as the control of COM position. When the COM motion state (i.e., its position and velocity relative to the BOS) is within the FSR (Figure 2.1b, point A), a person can maintain body balance without changing the BOS. When the COM motion state is below the lower limit (or the backward balance loss limit) of the FSR (Figure 2.1b, point B), it lacks the forward momentum to transport the COM over its BOS. In this scenario, a person must take a backward recovery step to keep the body from falling backward (like in a slip). Conversely, when the COM motion state is above the upper limit (or the forward balance loss limit) of the FSR (Figure 2.1b, point C), the COM has excessive forward momentum that would carry the COM beyond the toe of the BOS. A forward recovery step is unavoidable to avert falling forward (like in a trip). Dynamic gait stability is determined as the shortest distance from the COM motion state to the lower limb of the FSR (Figure 2b, solid lines).

FSR-based dynamic gait stability is more accurate in describing perturbation-related falls than the traditional static stability theory, and it is therefore a comprehensive tool that can quantify a person's reaction to an external perturbation (Yang et al., 2009). Furthermore, dynamic gait stability has been broadly used in the literature to quantify fall risk in both young (Yang et al., 2009; Yang et al., 2018c) and older (Bhatt et al., 2011; Yang and Pai, 2014) adults during motor tasks such as sit-to-stand (Wang et al., 2011), standing (Ahn et al., 2024a; Simpkins et al., 2022a; Simpkins and Yang, 2023a), and walking (Yang et al., 2008b) either without or with slip perturbations (Yang et al., 2018a), trip perturbations (Wang et al., 2012), or external load carriage (Ahn et al., 2022a; Ahn et al., 2022b; Simpkins et al., 2023; Simpkins et al., 2022b; Yang et al., 2022a). Although previous work suggested that local (Lockhart and Liu, 2008) and orbital stability (Grabiner et al., 2008; Hamacher et al., 2011) can differentiate individuals at a higher fall risk from non-fallers, more recent work suggests that dynamic gait stability as defined by the FSR could be more closely related to fall risk during walking in older adults compared with other stability measures such as the margin of stability, Floquet multiplier, Lyapunov exponent, and various gait parameters (step length, width, time) (Yang and Pai, 2014). Therefore, FSR-based dynamic gait stability provides a validated, accurate, and comprehensive method to examine older adults' responses to standing-slip and gait-slip perturbations. In addition to healthy adults, the FSR has been applied to ballet dancers (Simpkins et al., 2022a; Simpkins and Yang, 2023a, b) and individuals with various medical conditions such as stroke (Kajrolkar et al., 2014), multiple sclerosis (Lin et al., 2020; Yang et al., 2019), Parkinson's disease (Bhatt et al., 2013; Mak et al., 2011), and obesity (Liu and Yang, 2017; Yang et al., 2017).

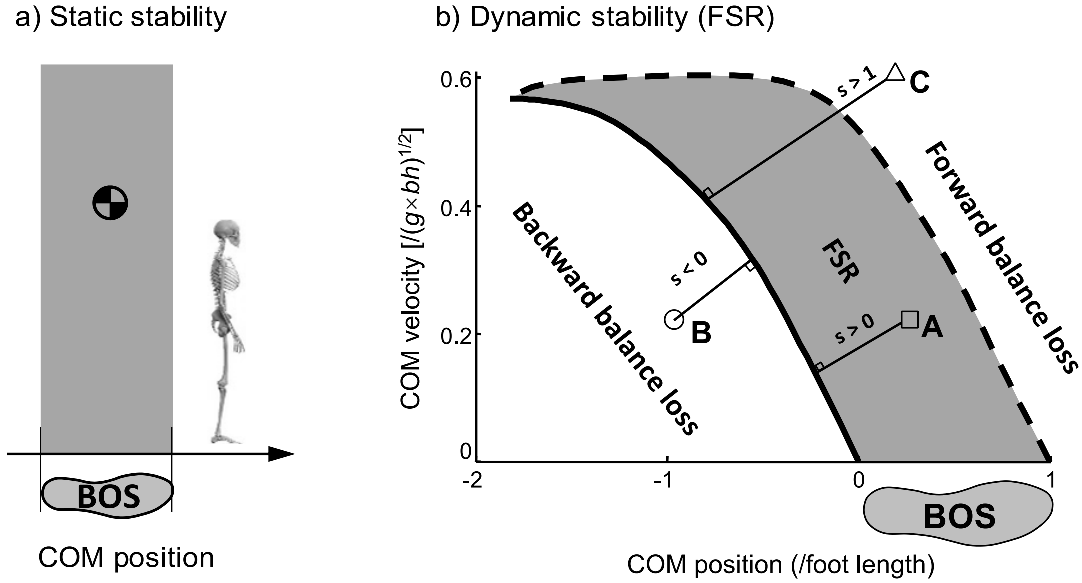

Figure 2.1 A depiction of a) the static stability limits and b) dynamic stability based on the Feasible Stability Region (FSR) theory (Yang, 2018). Static stability only considers the relative center of mass (COM) position to the base of support (BOS). Whenever the COM projection falls inside the BOS, a person is stable. The FSR considers both the COM position and velocity relative to BOS. The system's COM motion state's two components (i.e., its anteroposterior position and forward velocity) are calculated relative to the rear of the BOS (i.e., the leading heel) and normalized by foot length and $\sqrt{g \times bh}$, respectively, where g represents the acceleration due to gravity and bh the body height. When the COM motion state is above (point A)/below (point B) the FSR's backward balance loss limit, the dynamic stability is positive/negative, indicating a stable/unstable state against experiencing a backward fall. When the COM motion state is above the FSR's forward balance loss limit (point C), the dynamic stability is greater than one, indicating an unstable state against experiencing a forward fall. The dynamic gait stability value (s) is calculated as the shortest distance from the COM motion state to the backward balance loss limit of the FSR. Since both components of the COM motion state are normalized and dimensionless, dynamic gait stability is a unitless measurement.

3 METHODOLOGY

3.1 Study Design

This study adopted an observational case-control design with two groups for comparison: 1) healthy older adults who have taken ballet classes and 2) age- and sex-matched healthy older adults with no dance training experience. While on a special treadmill, each participant was exposed to a novel standing-slip followed by a series of fifteen additional repeated standing-slips mixed with non-slip trials. This standing-slip procedure was followed by three normal walking trials and one gait-slip trial. The slip response performance was compared between groups to test the hypotheses (Figure 3.1). The procedures for this project were approved by the Institutional Review Board at Georgia State University (Approval protocol number: H22622).

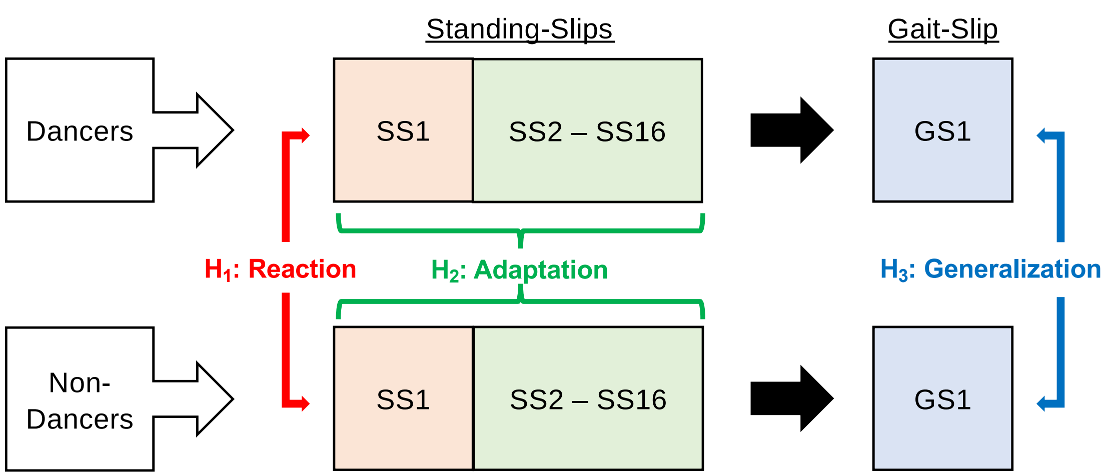

Figure 3.1 The schematic of the study design to test three research hypotheses. **Hypothesis 1** (H_1, reaction to the first standing-slip) states that ballet dancers will experience fewer falls with higher stability, a more effective recovery step, better controlled trunk movement, and shorter leg muscle electromyography (EMG) latencies after the first unexpected standing-slip (SS1) compared to non-dancers. **Hypothesis 2** (H_2, adaptation to repeated standing-slips) states that dancers will show a quicker adaptation to repeated standing-slip perturbations (SS2-SS16) than non-dancers. Specifically, dancers will improve their slip-reactions (including slip-faller rate, stability, trunk movement, and EMG latency) faster than the non-dancers. **Hypothesis 3** (H_3, generalization to novel gait-slip) states that ballet dancers will show more transfer of the fall resistant skills learned from standing-slips (SS1-SS16) to the novel gait-slip (GS1) than the non-dancers. In detail, dancers will experience fewer falls with higher stability, a more effective recovery step, better controlled trunk movement, and shorter leg muscle EMG latencies after the novel gait-slip compared to non-dancers.

3.2 Participants

Eligibility criteria for this study were 1) aged 55 or older, 2) no known acute or chronological neurological or musculoskeletal disorders, 3) no lower extremity fractures in the preceding three months, 4) no previous perturbation training experience, and 5) no severe cognitive impairment to ensure accurate recall of their fall history and their ability to follow instructions throughout the study. Recruitment to the ballet dancer group required current participation in at least one ballet class (of 60+ minutes) a minimum of one day a week for at least the previous three months. Recruitment to the non-dancer group required no previous formal dance training. The dancers were recruited from dance schools offering adult ballet classes in the greater Atlanta area. Recruitment for both groups was accomplished through word of mouth, e-mails, and fliers (on bulletin boards and posted on social media).

Twenty-five older ballet dancers and 38 non-dancers were screened for eligibility (Figure 3.2). After excluding five dancers and 15 non-dancers, 20 older adult ballet dancer participants and 23 age- and sex-matched non-dancers were recruited for the study (Figure 3.2). The age- and sex-matching process involved the recruitment of ballet dancer participants first, followed by the identification of non-dancer participants of comparable age and sex. The mean age of both groups was carefully tracked throughout the 17-month data collection process (July 2022 to November 2023) to ensure that the ages of the dancers and non-dancers were not significantly different. Informed consent was obtained from all participants before data collection following a full explanation of the purpose, procedure, and potential benefits and risks of the study. Participants were given adequate time to ask any questions about the study and its protocol before providing their written consent.

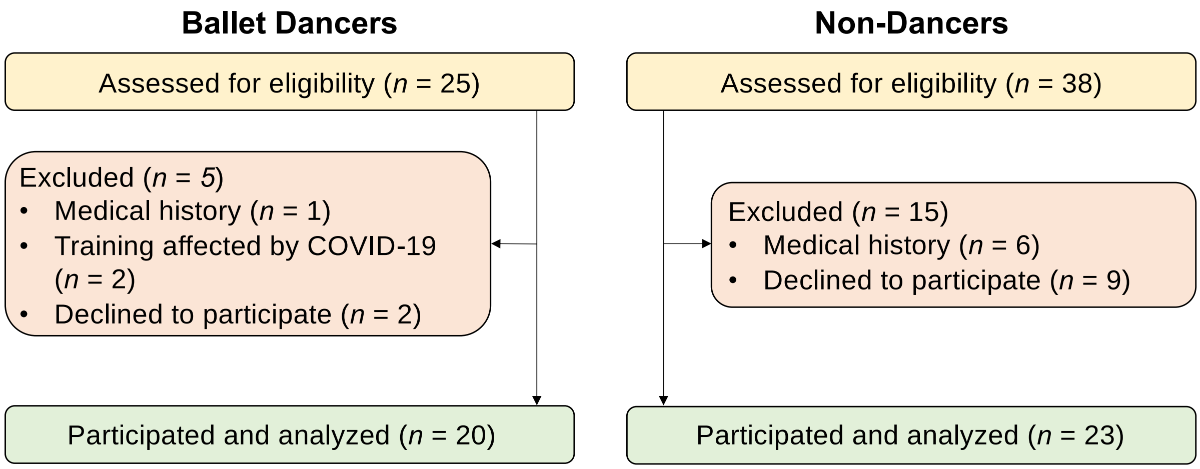

Figure 3.2 Flow chart of participant screening and recruitment.

3.3 Instruments

Motion Capture System

Kinematic data were collected using a 9-camera VICON motion capture system (VICON, Denver, CO, USA) sampling at a frequency of 100 Hz. All cameras were calibrated before each session to ensure data collection accuracy. To control the potential confounding effect of differing footwear, all subjects removed their shoes and were barefoot for the entire data collection (except for the Biodex strength assessment trials). Following anthropometric measurements, 26 reflective markers were applied to participants' bony landmarks according to the Helen-Hayes marker set: vertex, ears, rear neck, shoulders, right scapula, elbows, wrists, sacrum, greater trochanters, mid-thighs, knees, tibias, ankles, heels, and toes (Tabakin, 2000). One additional marker was placed on the treadmill belt for standing-slip trials to record the belt movement. Vicon Nexus 2.11 software (Oxford Metrics, Oxford, UK) was used during the data collection trials. All nine VICON cameras were connected to Vicon Nexus. The Vicon Nexus software was used to adjust the parameters, calibrate the cameras, and record the trials. All collected data were saved in Vicon Nexus.

ActiveStep Treadmill

The ActiveStep treadmill has been widely used to produce slip perturbations among various populations (Ahn et al., 2024a; Simpkins et al., 2022a; Simpkins and Yang, 2023a; Yang et al., 2013; Yang et al., 2019) because of its capacity to standardize the intensity of the perturbation (Simbex, NH). Such slip perturbations are typically induced by suddenly accelerating the belt speed forward during walking (Yang et al., 2019) or standing (Yang et al., 2018c). In this study, an ActiveStep treadmill was used for all standing-slip (Figure 3.3), non-slip, normal walking, and gait-slip (Figure 3.4) trials. Participants were fitted with a safety harness that was connected to a loadcell and then an overhead arch with ropes. The rope lengths were adjusted for each participant to ensure that only the feet contacted the treadmill belt and participants would not be able to step away from the treadmill base should a balance loss or fall occur.

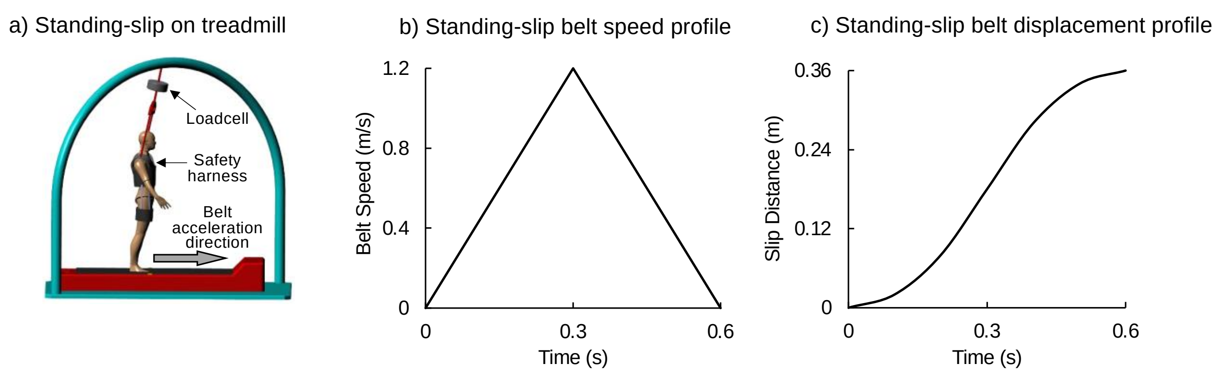

Figure 3.3 Schematics of a) the ActiveStep treadmill that was used to produce the standing-slip perturbation, and the profile of the b) speed and c) displacement of the treadmill belt for a standing-slip. The standing-slip perturbations were induced by quickly accelerating and then decelerating the treadmill belt over 0.6 s. The peak slip velocity was 1.2 m/s and the total slip distance was 0.36 m. The acceleration level was 4 m/s^2 (during the first or accelerating phase) or -4 m/s^2 (the second or decelerating phase). A safety harness protected participants during all trials on the treadmill.

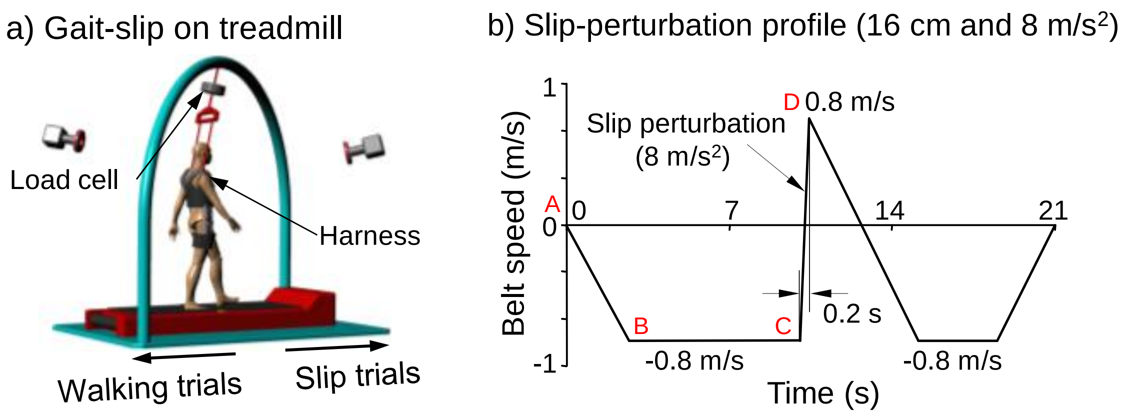

Figure 3.4 a) The ActiveStep treadmill that was used to induce the gait-slip and b) the profile of the treadmill belt speed during the gait-slip with an acceleration of 8 m/s^2 and a displacement of 0.16 m. The trial began with quiet standing on a stationary belt (A), and then the profile's initial gait speed was set at 0.8 m/s (B). The slip perturbation was created by suddenly and unexpectedly accelerating the belt speed from 0.8 m/s backward to 0.8 m/s forward (from C to D) within 200 ms after about 10 regular steps and approximately 80 – 120 ms after the lead foot touchdown.

Biodex Dynamometer

Maximum voluntary isometric contractions (MVIC) of the knee and ankle of the dominant leg were assessed using a Biodex Pro System 4 dynamometer (Biodex, NY, USA), sampling at a frequency of 100 Hz. Selection of the dominant leg was based on previous evidence that lower limb strength does not significantly differ between the dominant and non-dominant legs in healthy older adults (Ditroilo et al., 2010). Additionally, MVIC testing only on the dominant side reduced participants' burden and time. The MVICs included knee flexion, knee extension, ankle plantarflexion, and ankle dorsiflexion. The Biodex was equipped with handles to hold onto and a seatbelt for safety. The MVICs were performed three times per joint for each direction. Contractions had a duration of seven seconds with 30 seconds of rest in between contractions. The knee position during all knee flexion and extension trials was 35 degrees. The ankle position during the plantarflexion and dorsiflexion trials was 0 degrees. Participants wore socks and athletic shoes during Biodex testing. The maximum value of the three attempts was used to characterize the strength capacity of the respective muscle, which was normalized to the body mass (Nm/kg).

Electromyography System

Electromyography (EMG) analysis provides muscular activation information that is vital for understanding the biomechanics of human movement (De Luca, 1997). Analysis of EMG is also used to examine human balance recovery after a slip (Chambers and Cham, 2007; Marigold and Patla, 2002). The muscle activities of four lower extremity muscles were collected bilaterally during each trial using eight Delsys Trigno Wireless EMG sensors (Delsys Inc., Natick, MA, USA): rectus femoris (RF), biceps femoris (BF), tibialis anterior (TA) and medial gastrocnemius (MG). These muscles were selected based on previous work investigating lower extremity muscle activity in response to slip perturbations (Qu et al., 2012) and because these muscles are considered the major muscle groups in the lower extremity (Vaughan et al., 1992). These pairs of muscles also work as agonist-antagonist muscle groups in the thigh (RF/BF) and lower leg (TA/MG) (Qu et al., 2012). The EMG electrodes were placed over the belly of each muscle following the typical procedures (Ahn et al., 2022b), and the data were collected at a frequency of 1000 Hz (Figure 3.5). The EMG activity from each muscle was normalized to the MVIC of the dominant leg for each participant.

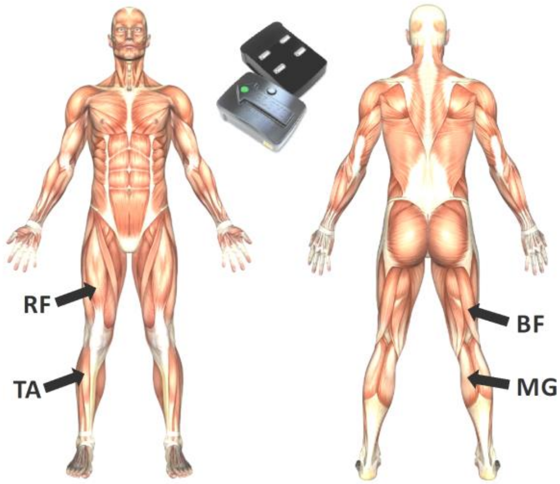

Figure 3.5 The electromyography (EMG) electrode placement. Eight EMG electrodes were placed bilaterally on the participants' rectus femoris (RF), biceps femoris (BF), tibialis anterior (TA), and medial gastrocnemius (MG).

3.4 Questionnaires and Physical/Cognitive Functions Assessment Tools

Questionnaires

1. Health and Dance History Questionnaire: A questionnaire was administered to all participants to collect basic demographic (age, sex, and gender), health, and physical activity information (Appendix A). All participants were asked to provide any information regarding previous medical diagnoses and to describe the type(s) of physical activity they currently engaged in. The dancer participants additionally completed a section of the questionnaire specific to their current and previous dance training.
2. Fall History Questionnaire: Fall history data for the previous 12 months were collected using a fall history questionnaire (Talbot et al., 2005). The questionnaire asked participants about any falls in the past 12 months and any fall-related injuries (Appendix B). This instrument has been used as the golden standard for collecting retrospective fall data in the literature (Talbot et al., 2005; Yang et al., 2023).

3. <u>Rapid Assessment of Physical Activity (RAPA) Questionnaire:</u> The RAPA Questionnaire (Topolski et al., 2006) collected physical activity level data for all participants (Appendix C). This questionnaire consists of two parts: 1) RAPA 1 for assessing aerobic activity and 2) RAPA 2 to evaluate strength and flexibility. Participants score themselves as "yes" or "no" on 10 items related to their weekly aerobic activity, strength training, and flexibility training. The RAPA 1 maximum score is 7 and uses the following scoring categories: a score of 0 = inactive, 1 = sedentary, 2 = under-active, 3 = under-active regular – light activities, 4 or 5 = under-active regular, and 6 or 7 = active. The highest score for RAPA 2 is 3, where strength training is worth 1 point and flexibility training is worth 2 points. Thus, the total maximum score for the entire RAPA test is 10, where a 0 indicates an inactive person and a 10 indicates a highly active person.

Montreal Cognitive Assessment

The Montreal Cognitive Assessment (MoCA) assesses short-term memory, visuospatial abilities, executive functions, attention, concentration, working memory, language, and orientation to time and place (Nasreddine et al., 2005). MoCA was given to participants by a certified MoCA administrator. The maximum MoCA score is 30 (a 0 indicates severe cognitive impairment a 30 indicates normal cognition).

Timed-Up-and-Go Test

All participants completed the Timed-Up-and-Go (TUG) test to assess dynamic balance and fall risk (Podsiadlo and Richardson, 1991). In the TUG test, participants began seated in a chair (seat height = 46 cm), were asked to rise without using the chair armrests, walk to a line on the floor (3 meters ahead of them), turn around, return to the chair, then sit back down "at a normal and comfortable pace." The TUG score was the time taken (in seconds) to complete the task.

Five-Time Sit-to-Stand Test

The Five-time Sit-to-Stand (5×STS) test can assess a person's fall risk and their leg muscle power (Goldberg et al., 2012; Simpkins and Yang, 2022b; Whitney et al., 2005; Zhang et al., 2013). Participants completed five cycles of rising from a chair (seat height = 46 cm) and sitting back down as fast as possible with arms folded across the chest (Goldberg et al., 2012). The time in seconds to finish this task was recorded.

3.5 Procedures

All data collection sessions took place in the Biomechanics Laboratory at Georgia State University, Atlanta, Georgia. Before participant arrival, all instruments and equipment were calibrated to ensure data accuracy.

Participant Preparation

Upon arrival, participants received a verbal explanation of the study procedures and were presented with the informed consent form. If the participants confirmed their interest in continuing with study participation, then the MoCA test was administered to ensure there would be no issues with their understanding of the instructions during the data collection session. If the participant met the minimum MoCA score of 25 for inclusion in this study (Pinto et al., 2019), then they proceeded to complete the following questionnaires:

1. Health and Dance History Questionnaire (Appendix A)
2. Fall History Questionnaire (Appendix B)
3. RAPA Questionnaire (Appendix C)

Following the questionnaires, participants completed the TUG test and the 5×STS test. Next, they underwent a five-minute walking warm-up and familiarization session on the ActiveStep treadmill. Then, basic anthropometric measurements were taken including body height, body mass, knee width, ankle width, ankle height, and the distance between the anterior superior iliac spines. Twenty-six reflective markers were attached to participants' bony landmarks using double-sided tape, and eight EMG electrodes were placed bilaterally on the RF, BF, TA, and MG. Before EMG electrode placement, all sites were shaved (if needed) and cleaned with rubbing alcohol to ensure a strong signal. After EMG electrode placement, the sensors were wrapped with pre-wrap and tape to secure them throughout the duration of the data collection.

Data Collection

For standing-slip trials, one additional reflective marker was placed on the treadmill belt to record its movement and identify the ON instant. Participants were fitted with a safety harness and stepped onto the treadmill. The harness was connected to a loadcell and then an overhead arch through dynamic ropes (Figure 3.3a). The rope lengths were adjusted to ensure that the harness would safely protect participants in case of balance loss while not interfering with their gait or posture on the treadmill.

Participants were told that for the first three standing trials (ST1-ST3, Figure 3.6), they would be performing normal standing (10 seconds per trial). Next, participants were informed that for the remaining standing trials, they "may or may not experience a 'slip-like' movement on the treadmill" without knowing when and how it would occur (Figure 3.6). The next three standing trials with a slip possibility did not actually involve a slip (ST4-ST6, Figure 3.6). Then, the novel and unexpected standing-slip (SS1) occurred. This was followed by mixed blocks of repeated slip (SS2-SS16) and non-slip (NS1-NS11) trials (Figure 3.6).

The standing-slips were induced by quickly accelerating the treadmill belt forward during quiet standing from 0.0 m/s to 1.2 m/s over 0.30 seconds, and then decelerating the belt speed to zero over another 0.30 seconds (Figure 3.3b). The belt slip distance was 0.36 meters (Figure 3.3c). Following the standing-slip procedure, participants completed three 10-second normal walking trials (Figure 3.6). The walking speed was set at 0.8 m/s for all participants. Subsequently, participants were exposed to an unexpected gait-slip during treadmill walking at 0.8 m/s (GS1; Figure 3.4). The belt acceleration, peak velocity, and slip displacement for the gait-slip were standardized at 8 m/s^2, 1.6 m/s, and 0.16 meters, respectively (Ding and Yang, 2016).

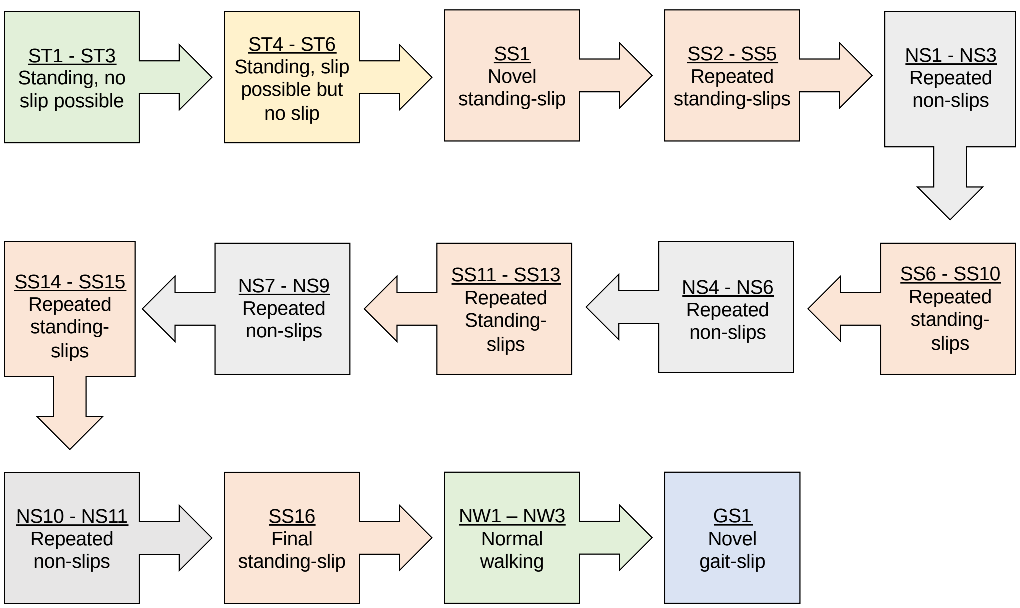

Figure 3.6 A flow chart of the standing-slip and gait-slip procedures. Each participant began on the ActiveStep treadmill with three standing trials (ST1-ST3) where no slip perturbation was possible. This was followed by three standing trials (ST4-ST6) where participants were aware that a slip "may or may not occur," but no slips occurred. Next, the novel slip (SS1) occurred followed by mixed blocks of standing-slips (SS2-SS16) and standing non-slips (NS1-NS11). After the standing-slip portion, participants walked normally on the treadmill for three trials (NW1-NW3) initially and then were exposed to a novel gait-slip (GS1).

Following the slip procedure, participants received a 20-minute rest period during which the safety harness, reflective markers, and non-dominant leg EMG electrodes were removed. As the final step of the data collection, the MVIC of the dominant knee (flexion and extension) and ankle (plantarflexion and dorsiflexion) was performed with participants seated on an isokinetic dynamometer (Biodex System 4, NY) to collect their isometric strength capacity and dominant leg EMG signals. During the knee joint trials, the participants' trunk and dominant thigh were stabilized with safety belts to minimize any unwanted movement. The rotational axis of the dynamometer was aligned with the transverse knee-joint axis, and the position of the knee was fixed at 35 degrees. For the knee joint strength assessments, participants were instructed to straighten (during extension) or bend (during flexion) their knees as hard as possible against the dynamometer during each contraction. There were three MVIC trials consisting of knee extension alternating with knee flexion. For the ankle joint strength assessments, participants were instructed to push against (during plantarflexion) or pull back against (during dorsiflexion) the dynamometer as hard as possible. They were also told to try not to engage the knee and hip joints during the plantarflexion attempts. The rotational axis of the dynamometer was aligned with the transverse ankle-joint axis, and the position of the ankle was fixed at 0 degrees. There were three MVIC trials consisting of ankle plantarflexion alternating with ankle dorsiflexion. For all MVIC trials, seven-second contractions were separated by a 30-second rest.

3.6 Data Processing and Reduction

Full-body kinematics were recorded during all slip and non-slip trials using the motion capture system. Bilateral leg muscle activity was collected using the Trigno wireless EMG system. The loadcell gathered the force applied to the safety harness. The belt's displacement in each standing-slip trial was registered by the marker on the treadmill belt. The motion capture, EMG signal, and loadcell force were synchronized through the Vicon Nexus system.

Outcome Measurements

Marker paths were low-pass filtered at marker-specific cutoff frequencies (ranging 4.5–9 Hz) using fourth-order, zero-lag Butterworth filters (Winter, 2005). Joint center, heel, and toe locations were computed from the filtered marker positions (Vaughan et al., 1992). For the standing-slips, the slip onset (ON) was identified as the instant when the anteroposterior position of the belt marker was three standard deviations above its baseline average (Figure 3.7a). The first recovery step for all standing-slip trials was the initial backward step taken after ON. The two transitional events of the recovery step liftoff (LO, Figure 3.7b) and its touchdown (TD, Figure 3.7c) were identified from the foot kinematics for the recovery step (Ahn et al., 2024a; Simpkins et al., 2022a; Simpkins and Yang, 2023a). The gait-slip ON was defined as the moment when the belt started increasing its speed from its steady value of 0.8 m/s backward (i.e., point C in Figure 3.4). For the gait-slip recovery analysis, the lead foot touchdown (LTD, Figure 3.8a), the recovery foot liftoff (RLO, Figure 3.8b), and the recovery foot touchdown (RTD, Figure 3.8c) were determined from the foot kinematics (Zeni Jr et al., 2008). All touchdown and liftoff events were verified by video recordings.

a) Slip onset　　b) Recovery step liftoff　　c) Recovery step touchdown

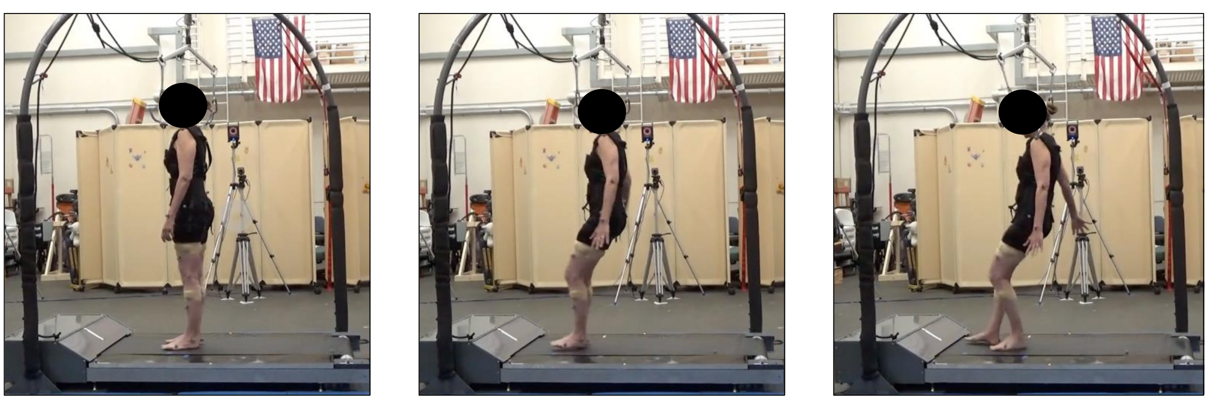

Figure 3.7 Images showing the three critical events of a) slip onset, b) recovery step liftoff, and c) recovery step touchdown during a standing-slip trial on the ActiveStep treadmill.

a) Lead foot touchdown b) Recovery foot liftoff c) Recovery foot touchdown

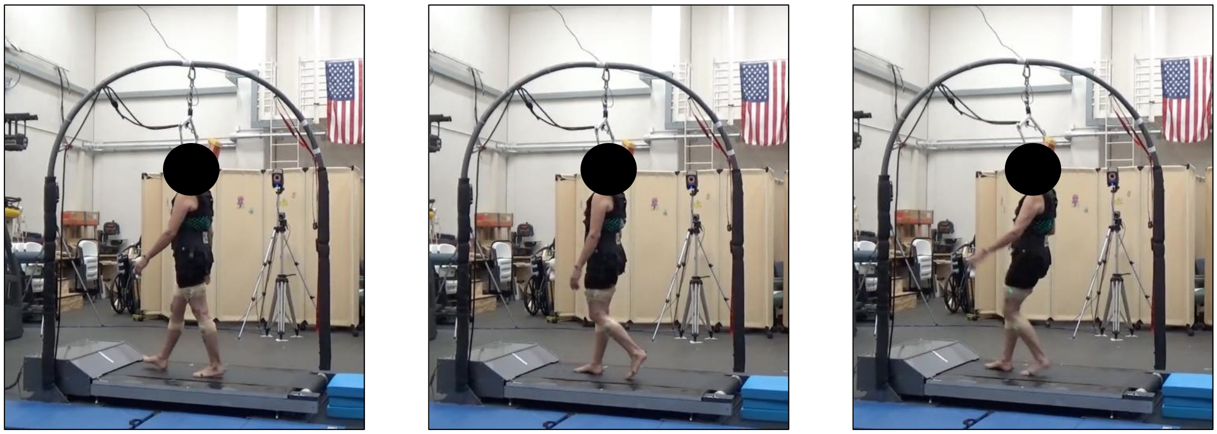

Figure 3.8 The three events of a) lead foot touchdown, b) recovery foot liftoff, and c) recovery foot touchdown during a gait-slip trial on the ActiveStep treadmill.

The body's COM kinematics were computed from joint centers using sex-dependent segmental inertial parameters (Yang et al., 2019). The COM motion state's two components (position and velocity) were calculated relative to the rear of the BOS and normalized by the foot length and $\sqrt{g \times bh}$, respectively, where g is the gravitational acceleration and *bh* the body height (Yang, 2018). Dynamic gait stability (primary outcome measure, continuous and unitless) was calculated for standing-slip trials (at ON, LO, and TD) and the gait-slip trial (at LTD, RLO, and RTD) using the COM motion state according to the FSR (Yang, 2018). The larger the stability value, the more stable a person is in resisting slip-related backward falling (Yang, 2018).

Secondary outcomes for standing-slip trials and the gait-slip trial included step length, step latency, step duration, step speed, slip distance, trunk angle, trunk angular velocity, slip outcome, and the slip-faller rate (Table 3.1). The recovery step length was determined as the anteroposterior distance between heels at TD for standing-slips (or RTD for the gait-slip) and normalized to *bh*. Step latency was the time between ON and LO (only calculated for standing-slip trials). Step duration was the interval between LO and TD for standing-slips (or between RLO and RTD for the gait-slip). Step speed was the step length divided by its duration. Slip distance was the treadmill belt marker's displacement between ON and LO for a standing-slip (or the leading foot's displacement from the ON to its most anterior position for the gait-slip).

The trunk angle in degrees at ON, LO, and TD (or LTD, RLO, and RTD for the gait-slip) was calculated as the angle formed by the trunk segment and the vertical reference line. A 0-deg trunk angle indicates an upright trunk and negative trunk angle values represent a backward leaning trunk. The instantaneous trunk angular velocity (in deg/s) was calculated at ON, LO, and TD (or LTD, RLO, and RTD for the gait-slip) as the first-order derivative of the trunk angle with respect to time. Slip outcomes (fall vs. non-fall) were classified as falls if the peak loadcell force exceeded 30% of the bodyweight (Yang and Pai, 2011) and were verified with video recordings. The slip-faller rate on each slip trial was the ratio of the number of fallers to the total participant number in each group. The primary and secondary outcome measures at ON for a standing-slip and LTD for the gait-slip were utilized to identify proactive adjustments during repeated standing-slip adaptation, whereas the primary and secondary outcome measures at LO and TD for standing-slip and RLO and RTD for the gait-slip were used to identify reactive adjustments.

The EMG signals were filtered by a band-pass filter between 20-500 Hz, rectified, and filtered with a low-pass filter with a cutoff frequency of 10 Hz (Ahn et al., 2022b). Processed EMG signals were normalized and presented as a percentage of the MVIC. EMG latency (in seconds) was calculated for all standing-slip trials as the time elapsed from the onset of the belt's slip perturbation to the onset of an EMG burst exceeding the magnitude of baseline level in the respective muscle (Simmons, 2005; Smith et al., 1996). The EMG latency for the gait-slip trial was calculated as the time elapsed from the slip onset to the EMG onset, where the EMG onset was determined using a preset threshold and then corrected as needed following visual inspection

(Nieuwboer et al., 2004). Due to technical difficulties, the EMG signals were not usable for some trials and were excluded from the final analyses.

Table 3.1 The secondary outcome variables and their definitions for both a standing-slip and a gait-slip. Some variables are defined based on certain events such as the recovery foot liftoff (LO or RLO), recovery step touchdown (TD or RTD), or the lead foot touchdown (LTD). For the gait-slip trial, the underlined portion in the second column is replaced by the content provided in the third column of the table.

Outcome	**Definition of secondary variables**	
	Standing-Slip	**Gait-Slip**
Step Length	Anteroposterior distance between heels at TD, normalized to *bh*	RTD
Step Latency	Time between slip onset and LO	
Step Duration	Interval between LO and TD	RLO and RTD
Slip Distance	Treadmill belt marker's displacement between slip onset and LO	The most anterior position of the leading foot
Trunk Angle	Angle formed by trunk segment and the vertical reference line at slip onset, LO, and TD	LTD, RLO, and RTD
Trunk Angular Velocity	First-order time derivative of the trunk angle at slip onset, LO, and TD	LTD, RLO, and RTD
Step Speed	Step length divided by step duration	
Slip Outcome	Fall vs non-fall, classified as falls if the peak loadcell force exceeds 30% of the bodyweight	
Slip-Faller Rate	Ratio of number of fallers to the sample size in each group	
EMG Latency	Time between slip onset and EMG burst onset	

3.7 Statistical Analyses

Normality and homogeneity of variance were checked prior to analyses for all variables with Shapiro-Wilk and Levene's tests, respectively. Variables that were not normally distributed were transformed as needed depending on the skew direction. For variables that continued to violate the normality assumption following transformation or violated Levene's test, the non-parametric Mann-Whitney *U* test was used for statistical analysis. In the case of a significant difference between groups in a demographic variable or the physical activity level, correlation analyses were performed, and the analysis of covariance (ANCOVA) model was utilized as needed. Independent *t*-test effect sizes were calculated as Cohen's *d*. The original effect sizes for χ^2 or Fisher's exact test (φ), Mann-Whitney test ($r = \frac{Z-statistic}{\sqrt{sample\ size}}$) and ANCOVA (partial η^2) were converted to Cohen's *d*. Effect sizes were interpreted as small ($0.2 \leq d < 0.5$), medium ($0.5 \leq d < 0.8$), or large ($d \geq 0.8$) (Cohen, 2013). The 95% confidence intervals (95% CI) for Cohen's *d* were also computed.

To test the first hypothesis, independent *t*-tests compared primary and continuous secondary outcome measures (χ^2 test for the slip outcome) between ballet dancers and non-dancers for the novel standing-slip trial (Figure 3.1). To test the second hypothesis, the percent change from the first slip to the last slip was calculated for all primary and secondary continuous variables and compared between groups using independent *t*-tests (Figure 3.1). A multiple linear regression analysis compared the change in the faller rate from the first slip to the last slip between groups. In the regression model, the log-transformed faller rate was the dependent variable while the trial number and group were the independent variables. A significant trial number by group interaction effect would indicate differences in the change in the faller rate between groups. The third hypothesis was tested by comparing primary and continuous secondary outcome measures (Fisher's exact test for the slip outcome) between groups for the novel gait-slip (GS1) using independent *t*-tests or appropriate non-parametric substitutions (Figure 3.1). All statistical analyses were conducted using SPSS 29.0 (IBM, NY) with an α of 0.05.

4 RESULTS

4.1 Participant Demographics

Independent *t*-tests revealed no significant group-related differences for age (p = 0.097, d = 0.520, 95% CI: [-0.093, 1.127]), sex (p = 1.000, d = 0.065, 95% CI: [-0.537, 0.662]), and body height (p = 0.325, d = 0.304, 95% CI: [-0.300, 0.905], Table 4.1). Dancers had a significantly lower body mass than non-dancers (p = 0.014, d = 0.789, 95% CI: [0.161, 1.407], Table 4.1). Dancers displayed a higher total physical activity score (p < 0.001, d = 1.366, 95% CI: [0.513, 1.814]) but a similar fall history (p = 0.704, d = 0.174, 95% CI: [-0.432, 0.769], Table 4.1) compared to the non-dancers. The RAPA 1 mean scores classified the dancers as "active" and the non-dancers as "under-active regular," and these scores were statistically different between groups (p = 0.006, d = 0.915, 95% CI: [0.296, 1.560], Table 4.1). The RAPA 2 scores were also higher in dancers than non-dancers (p = 0.006, d = 0.928, 95% CI: [0.315, 1.582], Table 4.1).

Table 4.1 Comparisons of demographic information (in mean ± standard deviation for continuous ones and count (%) for binary ones) for ballet dancers versus non-dancers. Physical activity level (aerobic, strength and flexibility, and total) and fall history are also provided. The maximum score of the total measured physical activity level is 10 (a 0 indicates an inactive person and a 10 indicates a highly active person). Independent *t*-tests compared age, body height, and body mass between groups. Fisher's exact test compared sex and fall history between groups. Mann-Whitney *U* test analyzed the aerobic activity scores, strength and flexibility scores, and total physical activity level scores. Effect sizes (Cohen's *d*) are also provided.

Parameter	**Group**		***p*-value**	**Effect size *(d)***
	Dancers (n = 20)	Non-Dancers (n = 23)		
Age (years)	63.85 ± 7.60	67.48 ± 6.40	0.097	0.520
Height (m)	1.62 ± 0.08	1.65 ± 0.10	0.325	0.304
Mass (kg)	61.39 ± 11.28	72.29 ± 15.71	0.014	0.789
Sex (female, %)	17 (85%)	19 (83%)	1.000	0.065
Physical activity total (/10)	9.30 ± 0.86	7.26 ± 2.24	< 0.001	1.366
Aerobic activity (/7)	6.45 ± 0.69	5.30 ± 1.55	0.006	0.915
Strength and flexibility (/3)	2.85 ± 0.37	2.00 ± 1.17	0.006	0.928
Fallers in past year (%)	3 (15%)	5 (22%)	0.704	0.174

4.2 Ballet Dancer Training and Experience

The 20 ballet dancers were currently training in ballet classes at various dance schools around the metropolitan Atlanta area on 2.25 ± 1.25 days per week for 81.00 ± 14.10 minutes per ballet class. The duration for maintaining this ballet training schedule varied widely between the dancers and ranged from six months to over 12 years. In addition to ballet, the dancers also participated in the following forms of physical activity: walking, strength training, yoga, running, cycling, hiking, gymnastics, swimming, and Zumba. The frequency of weekly participation in these additional physical activities varied from once a week to daily.

4.3 Non-Dancer Physical Activity

The 23 non-dancer participants were currently participating in the following forms of physical activity: walking, Pilates, golf, strength training, yoga, hiking, tai chi, cycling, running, rowing, pickleball, Zumba, high-intensity interval training, aerobics, and water aerobics. The frequency of weekly participation in these activities varied from once a week to daily.

4.4 Cognitive and Physical Functions

The MoCA score, TUG time, and 5×STS time all violated normality and could not be corrected via transformation. These three variables were analyzed using Mann-Whitney *U* tests. MoCA scores were not statistically different between groups ($p = 0.205$, $d = 0.404$, 95% CI: [-0.182, 1.031], Table 4.2). Dancers performed better than non-dancers for completing both the TUG ($p = 0.003$, $d = 1.033$, 95% CI: [0.235, 1.524]) and 5×STS ($p < 0.001$, $d = 1.280$, 95% CI: [0.546, 1.887], Table 4.2). All variables for leg strength were analyzed using independent *t*-tests. Compared to non-dancers, dancers displayed significantly stronger knee extensors ($p = 0.010$, $d = 0.821$, 95% CI: [0.192, 1.442]) and ankle plantarflexors ($p = 0.031$, $d = 0.689$, 95% CI: [0.061, 1.309], Table 4.2). No significant group differences were detected for knee flexors (*p* = 0.398, *d* = 0.261, 95% CI: [-0.342, 0.862]) or ankle dorsiflexors (*p* = 0.227, *d* = 0.234, 95% CI: [-0.375, 0.840], Table 4.2).

Table 4.2 Comparisons of cognition, physical function, and leg muscular strength (in mean ± standard deviation) for ballet dancers versus non-dancers. Mann-Whitney *U* tests compared the Montreal Cognitive Assessment (MoCA) scores, the Timed-Up-And-Go (TUG) test, and the Five-Time Sit-to-Stand (5×STS) test. Independent *t*-tests compared strength variables. Effect sizes in Cohen's *d* are also presented.

Parameter	**Group**		***p*-value**	**Effect size (*d*)**
	Dancers (n = 20)	Non-Dancers (n = 23)		
MoCA* (/30)	28.47 ± 1.58	27.73 ± 1.88	0.205	0.404
TUG* (sec)	9.59 ± 2.11	11.48 ± 2.16	0.003	1.033
5×STS* (sec)	11.41 ± 2.18	14.57 ± 2.90	< 0.001	1.280
Knee extensor strength (Nm/kg)	1.54 ± 0.37	1.27 ± 0.28	0.010	0.821
Knee flexor strength (Nm/kg)	1.07 ± 0.31	0.99 ± 0.25	0.398	0.261
Ankle plantarflexor strength# (Nm/kg)	1.07 ± 0.24	0.86 ± 0.35	0.031	0.689
Ankle dorsiflexor strength# (Nm/kg)	0.35 ± 0.07	0.34 ± 0.08	0.227	0.234

*: Missing data for one dancer and one non-dancer participant.
#: Missing data for one non-dancer participant.

4.5 Hypothesis 1: Reaction

Seven variables violated normality. Two variables (stability at LO and step latency) were transformed to successfully correct the abnormality using the logarithmic base 10 and reciprocal transformations, respectively. The normality for five variables (step duration, step length, step speed, trunk angle at LO, and MG standing latency) could not be corrected by transformation, therefore these variables were analyzed using Mann-Whitney *U* tests. Additionally, four variables violated Levene's test (stability at ON, COM position at ON, COM velocity at ON, and trunk velocity at ON) and were also analyzed using Mann-Whitney *U* tests. Three variables (stability at LO, trunk angle at ON, and trunk angle at TD) were significantly correlated with body mass, therefore these variables were analyzed using ANCOVA with body mass as the covariate.

No variables were significantly correlated with physical activity level. To reduce Type I error due to multiple comparisons, Holm's Sequential Bonferroni correction was made to the significance level for the primary outcome variables (Holm, 1979). The adjusted significance level varied from 0.017 (0.05/3) to 0.05 (0.05/1) for the primary outcome variables. All effect sizes are presented as Cohen's *d* with its 95% CI.

4.5.1 Primary outcome – Dynamic gait stability

Following the first unexpected standing-slip trial, dynamic gait stability was similar between groups at ON (p = 0.052, d = 0.514, 95% CI: [0.015, 1.244], Figure 4.1a), and significantly higher in dancers at LO (p = 0.006, d = 0.830, 95% CI: [0.168, 1.436], Figure 4.1b) and TD (p = 0.012, d = 0.722, 95% CI: [0.099, 1.337], Figure 4.1c, Tables A1-A2), where a higher dynamic gait stability value indicates a more stable state against backward balance loss resulting from the slip.

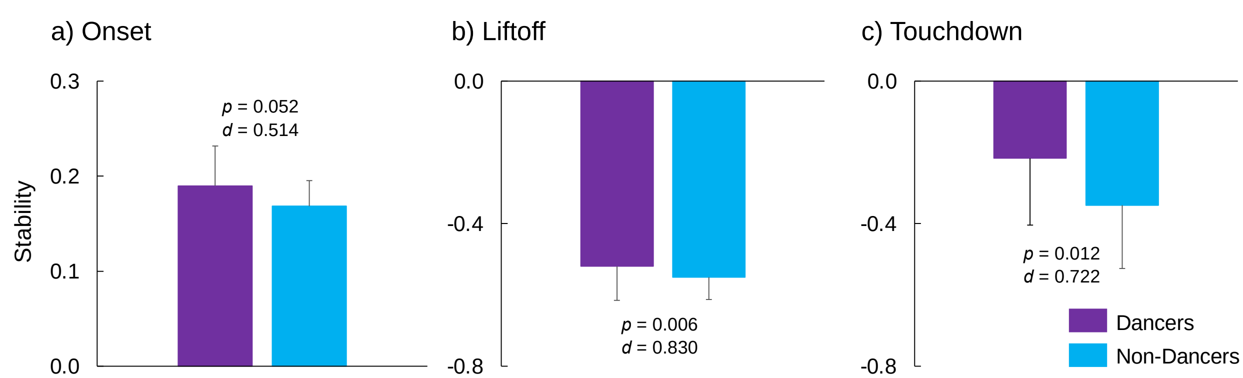

Figure 4.1 Comparisons of the dynamic gait stability at a) slip onset (ON), b) recovery step liftoff (LO), and c) recovery step touchdown (TD) between dancers (n = 20) and non-dancers (n = 23). Dynamic gait stability was calculated as the shortest distance from the COM motion state to the limit against backward balance loss (section 2.6). The effect size is provided as Cohen's *d*. The column height and the error bar represent the group mean and standard deviation, respectively.

4.5.2 Secondary outcomes

The COM position was significantly more anterior to the BOS in dancers than non-dancers at ON (p = 0.038, d = 0.563, 95% CI: [0.031, 1.263], Figure 4.2a), LO (p = 0.017, d = 0.670, 95% CI: [0.050, 1.282], Figure 4.2b), and TD (p = 0.012, d = 0.723, 95% CI: [0.099, 1.338], Figure 4.2c, Tables A1-A2). COM velocity was not significantly different between groups at ON (p = 0.090, d = 0.417, 95% CI: [-0.191, 1.021], Figure 4.2d) or TD (p = 0.428, d = 0.056, 95% CI: [-0.673, 0.526], Figure 4.2e), but the dancers had a significantly faster and more backward COM velocity at LO (p = 0.002, d = 0.944, 95% CI: [0.306, 1.572], Figure 4.2f, Tables A1-A2) than the non-dancers. For recovery stepping, no significant between-group difference was detected for the step length (p = 0.262, d = 0.195, 95% CI: [-0.436, 0.764], Figure 4.3a, Tables A1-A2). Compared to the non-dancers, dancers had a significantly shorter step latency (p = 0.020, d = 0.650, 95% CI: [0.031, 1.261], Figure 4.3b), a shorter step duration (p = 0.011, d = 0.747, 95% CI: [-0.019, 1.206], Figure 4.3c), and a faster step speed (p = 0.032, d = 0.590, 95% CI: [-0.088, 1.132], Figure 4.3d, Tables A1-A2). The slip distance at LO was significantly shorter in dancers than in non-dancers (p = 0.015, d = 0.694 [0.072, 1.308], Figure 4.3e, Tables A1-A2).

Trunk angle and trunk angular velocity were not significantly different between dancers and non-dancers at ON (angle: p = 0.140, d = 0.138, 95% CI: [0.000, 0.944], Figure 4.4a; velocity: p = 0.140, d = 0.335, 95% CI: [-0.210, 1.000], Figure 4.4d), LO (angle: p = 0.142, d = 0.330, 95% CI: [-0.128, 1.089], Figure 4.4b; velocity: p = 0.157, d = 0.312, 95% CI: [-0.293, 0.913], Figure 4.4e), or TD (angle: p = 0.344, d = 0.127, 95% CI: [0.000, 0.712], Figure 4.4c; velocity: p = 0.155, d = 0.315, 95% CI: [-0.290, 0.916], Figure 4.4e). However, the dancers consistently displayed a more upright (less backward leaning) trunk than non-dancers at all three instants (Tables A1-A2). Regarding the slip outcome, nine out of 20 dancers experienced a fall, and 19 out of 23 non-dancers experienced a fall following the novel standing slip trial. A χ^2 test compared the slip outcome between groups. The dancers displayed a significantly lower faller rate of 45% compared to the non-dancers' faller rate of 83% (p = 0.005, d = 0.970, 95% CI: [0.202, 1.735], Figure 4.5).

For EMG latency, the dancers exhibited significantly shorter muscle activity latencies than non-dancers for the standing side RF (p = 0.028, d = 0.661, 95% CI: [-0.016, 1.329], Figure 4.6a), stepping side BF (p = 0.031, d = 0.628, 95% CI: [-0.029, 1.277], Figure 4.6b), stepping TA (p = 0.017, d = 0.719, 95% CI: [0.054, 1.375], Figure 4.6c), standing TA (p = 0.002, d = 1.003, 95% CI: [0.335, 1.659], Figure 4.6c), and stepping MG (p = 0.030, d = 0.732, 95% CI: [-0.027, 1.481], Figure 4.6d, Tables A3-A4). The muscle latencies for the stepping RF (p = 0.233, d = 0.247, 95% CI: [-0.415, 0.905], Figure 4.6a), standing BF (p = 0.118, d = 0.377 [-0.245, 0.994], Figure 4.6b), and standing MG (p = 0.420, d = 0.076, 95% CI: [-0.890, 0.600], Figure 4.6d) were not statistically different between the two groups (Tables A3-A4). Due to technical difficulties and equipment malfunction, the EMG latencies for some participants could not be collected thus decreasing the sample size by various amounts for each variable (Figure 4.6).

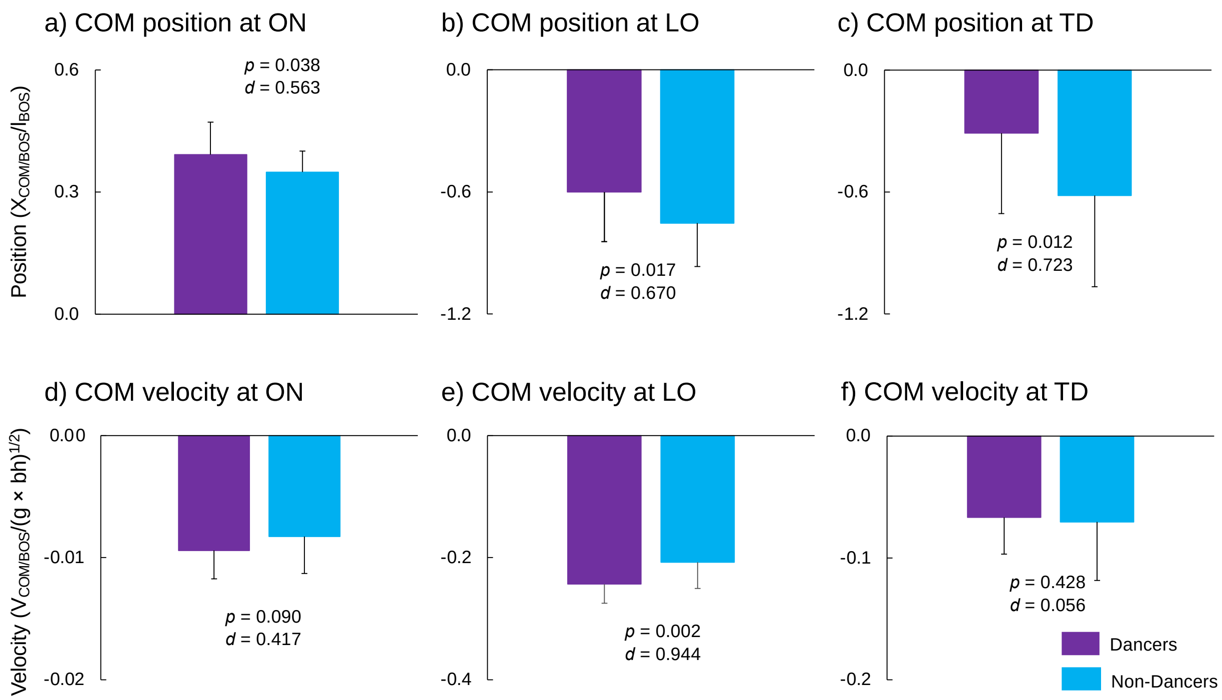

Figure 4.2 Comparisons of the center of mass (COM) position at a) slip onset (ON), b) recovery step liftoff (LO), and c) recovery step touchdown (TD), and the COM velocity at d) ON, e) LO, and f) and TD between ballet dancers (n = 20) and non-dancers (n = 23) on the first standing-slip. The COM position and velocity are relative to the rear of the base of support (BOS) and normalized by the foot length (l_{BOS}) and $\sqrt{g \times bh}$, where g is the gravitational acceleration and bh is the body height. Effect sizes (Cohen's d) are provided. The column height and the error bar represent the group mean and standard deviation, respectively.

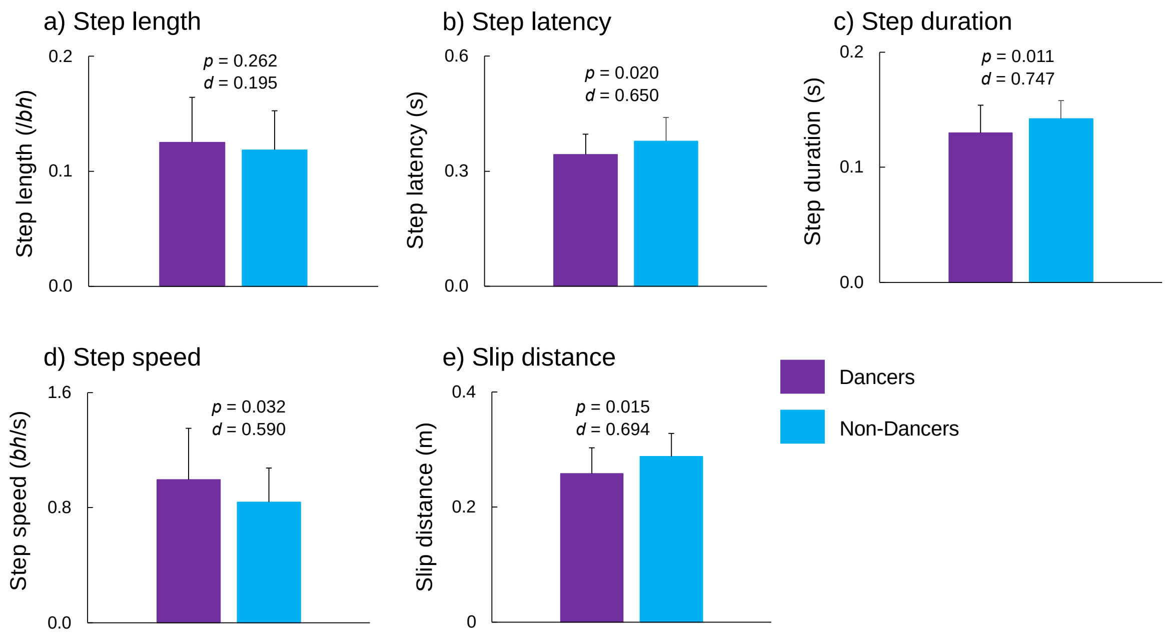

Figure 4.3 Comparisons of the recovery step a) length, b) latency, c) duration, d) and speed, and e) slip distance at recovery liftoff (LO) between dancers (n = 20) and non-dancers (n = 23) on the first standing-slip. Step length was the anteroposterior distance between heels at the recovery step touchdown (TD) and expressed in body height (*bh*). Step latency was the time elapsed between slip onset (ON) and LO. Step duration was the interval between LO and TD. Step speed was calculated as the recovery step length divided by its duration. Slip distance was the treadmill belt's displacement between ON and LO. Effect sizes (Cohen's *d*) are provided. The column height and the error bar represent the group mean and standard deviation, respectively.

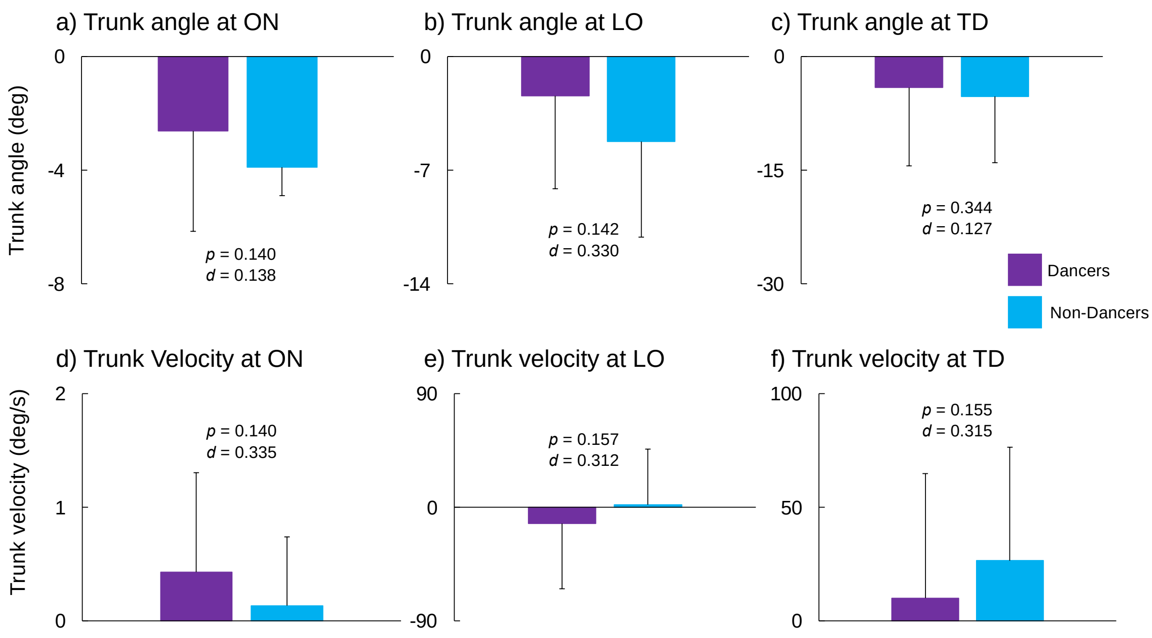

Figure 4.4 Comparisons of the trunk angle at the a) slip onset (ON), b) recovery step liftoff (LO), and recovery step touchdown (TD), and the trunk angular velocity at d) ON, e) LO, and f) TD between dancers (n = 20) and non-dancers (n = 23). The trunk angle in degrees was the angle formed by the trunk segment and the vertical reference line (where a negative angle indicates backward lean). The instantaneous trunk angular velocity (in deg/s) was the first derivative of the trunk angle with respect to time. Effect sizes are expressed as Cohen's d. The column height and the error bar represent the group mean and standard deviation, respectively.

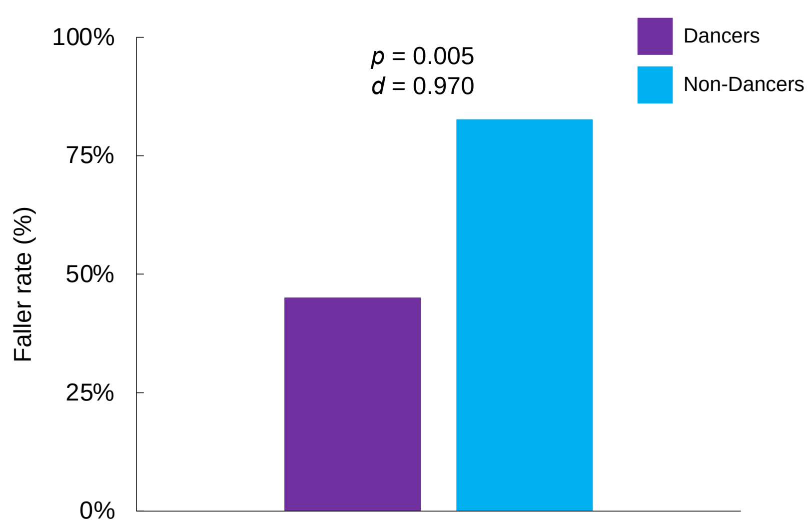

Figure 4.5 Comparison of the faller rate between dancers (n = 20) and non-dancers (n = 23). The slip outcome (fall vs. non-fall) was classified as a fall if the peak loadcell force exceeded 30% of the bodyweight and then was also verified by video recordings. The slip-faller rate was the ratio of the number of fallers to the total participant number in each group. The effect size (Cohen's d) is provided.

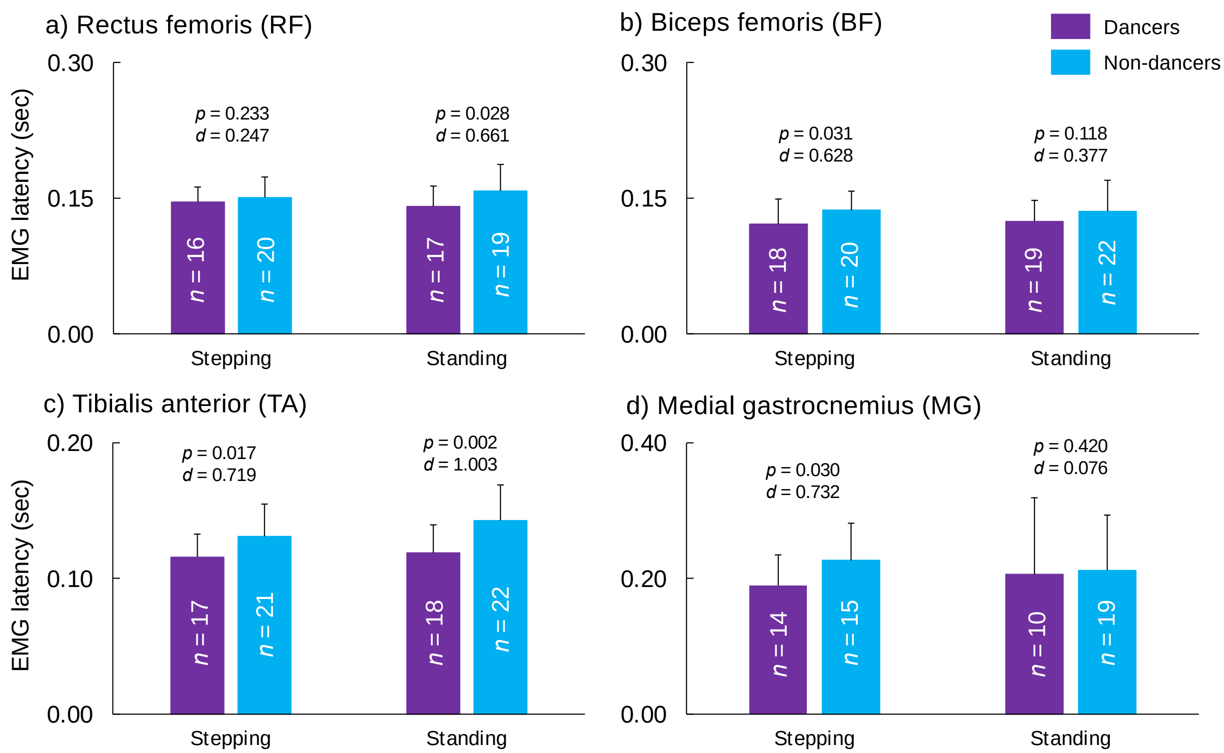

Figure 4.6 Comparisons of electromyography (EMG) latencies during the first standing-slip of the stepping and standing leg a) rectus femoris, b) biceps femoris, c) tibialis anterior, and d) medial gastrocnemius between dancers and non-dancers. EMG latency (sec) was the time from the onset of the belt's slip perturbation to the onset of an EMG burst exceeding the magnitude of the baseline level in the respective muscle. The effect sizes (Cohen's *d*) are provided. The column height and the error bar represent the group mean and standard deviation, respectively. The number inside each bar indicates the number of participants used for the comparisons. Due to technical difficulties, the EMG signals were not usable for some participants.

4.6 Hypothesis 2: Adaptation

Thirty-seven participants (15 dancers and 22 non-dancers) completed all 16 standing-slip trials. Four dancers and one non-dancer only completed 10 standing-slip trials (recruited to the project before the additional six standing-slips were added to the protocol), and one dancer only completed 13 standing-slip trials (requested to terminate the protocol early due to scheduling conflict). To quantify adaptation from the first to the last slip, the percent change from each participant's first to their last standing-slip trial was calculated for all variables and then averaged for each group. Sixteen of the percent change variables (stability at LO and TD, COM position at LO and TD, COM velocity at TD, step length, step speed, trunk angle at ON, LO, and TD, trunk velocity at ON, LO, and TD, slip distance, RF stepping latency, and TA stepping latency) violated normality. The violation was rectified for two variables (stability at LO and COM position at LO) using the power of two and exponential transformations, respectively. The remaining 14 variables could not be corrected and were analyzed using Mann-Whitney *U* tests. One variable (COM velocity at ON) violated Levene's test and was also analyzed using the Mann-Whitney *U* test. Three of the percent change variables (COM position at LO, COM velocity at LO, and step latency) were significantly correlated with body mass, and one variable (stability at LO) was significantly correlated with the total physical activity score. These variables were analyzed using ANCOVA with body mass or the total physical activity score as the covariate. To reduce Type I error due to multiple comparisons, Holm's Sequential Bonferroni correction was made to the significance level for the primary outcome variables (Holm, 1979). The adjusted significance level varied from 0.017 (0.05/3) to 0.05 (0.05/1) for the primary outcome variables. All effect sizes are presented as Cohen's *d* with 95% CI.

4.6.1 Primary Outcome: Change in Dynamic Gait Stability

From the first standing-slip to the last standing-slip, the percent change in dynamic gait stability was not significantly different between groups at ON ($p = 0.264$, $d = 0.194$, 95% CI: [-0.407, 0.794], Figure 4.7a) or TD ($p = 0.069$, $d = 0.464$, 95% CI: [-0.234, 0.975], Figure 4.7c, Tables A5-A6). At LO, the dancers improved stability significantly more than non-dancers ($p = 0.007$, $d = 0.720$, 95% CI: [0.138, 1.406], Figure 4.7b, Tables A5-A6).

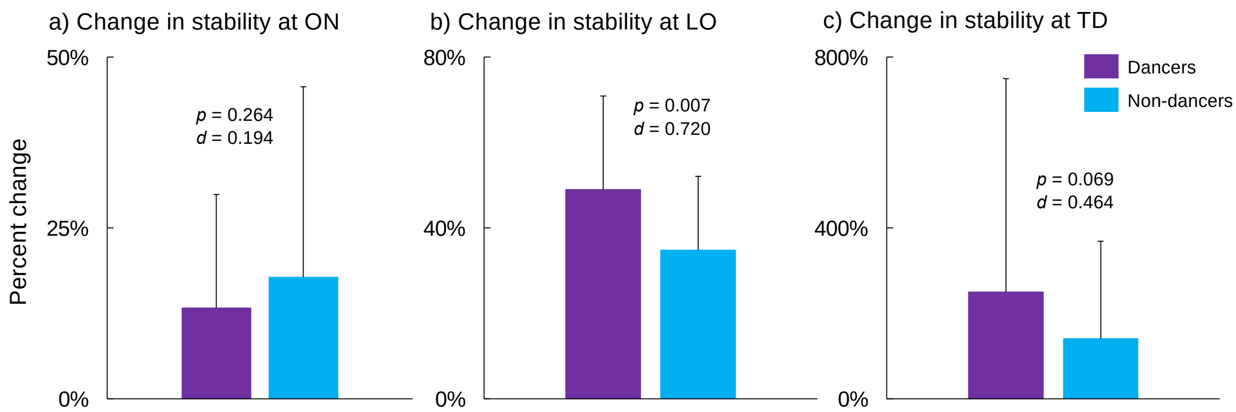

Figure 4.7 Comparisons of the percent change in dynamic gait stability from first standing-slip to last standing-slip at a) slip onset (ON), b) recovery step liftoff (LO), and c) recovery step touchdown (TD) between dancers (n = 20) and non-dancers (n = 23). Dynamic gait stability was calculated as the shortest distance from the COM motion state to the threshold against backward balance loss. Effect sizes (Cohen's d) are provided. The column height and the error bar represent the group mean and standard deviation, respectively.

4.6.2 Secondary Outcomes

The percent change in COM position from the first to last standing-slip was not significantly different between groups at ON (p = 0.232, d = 0.226, 95% CI: [-0.376, 0.826], Figure 4.8a), LO (p = 0.191, d = 0.278, 95% CI: [0.000, 0.879], Figure 4.8b), or TD (p = 0.326, d = 0.138, 95% CI: [-0.667, 0.531], Figure 4.8c, Tables A5-A6). The percent change in COM velocity from first to last slip was not different between groups at ON (p = 0.233, d =0.223, 95% CI: [-0.230, 0.980], Figure 4.8d), but it was significantly different between groups at LO (p = 0.001, d = 0.972, 95% CI: [0.370, 1.658], Figure 4.8e) and TD (p = 0.035, d = 0.577, 95% CI: [-0.025, 1.200], Figure 4.8f, Tables A5-A6). At LO, the dancers maintained a similar COM velocity from the first to the last slip, whereas the non-dancers exhibited a reduced COM velocity (or an increased velocity in the backward direction). At TD, the dancers were able to reverse their COM velocity from backward to forward from the first to the last slip, and the non-dancers still showed increased backward COM velocity. For the change in recovery stepping variables, no significant between-group differences were detected for step length (*p* = 0.486, *d* = 0.012, 95% CI: [-0.740, 0.460], Figure 4.9a) or step latency (*p* = 0.243, *d* = 0.220, 95% CI: [0.000, 0.820], Figure 4.9b, Tables A5-A6). The dancers increased their step duration more than the non-dancers (*p* = 0.004, *d* = 0.861, 95% CI: [0.229, 1.483], Figure 4.9c), whereas the non-dancers increased their step speed more than the dancers (*p* = 0.011, *d* = 0.755, 95% CI: [0.072, 1.308], Figure 4.9d, Tables A5-A6). The slip distance decreased more in dancers than non-dancers from first to last slip (*p* = 0.010, *d* = 0.769, 95% CI: [-0.123, 1.094], Figure 4.9e, Tables A5-A6).

There were no significant between-group differences for the percent change in trunk angle from the first to last slip at ON (*p* = 0.313, *d* = 0.148, 95% CI: [-0.426, 0.775], Figure 4.10a), LO (*p* = 0.233, *d* = 0.223, 95% CI: [-0.767, 0.434], Figure 4.10b), or TD (*p* = 0.064, *d* = 0.335, 95% CI: [-0.505, 0.694], Figure 4.10c, Tables A5-A6). For trunk angular velocity, there was no group difference at ON (*p* = 0.331, *d* = 0.134, 95% CI: [-0.691, 0.509], Figure 4.10d, Tables A5-A6). However, at LO the dancers improved their trunk velocity (or rotating forward) more than the non-dancers (*p* = 0.011, *d* = 0.750, 95% CI: [0.222, 1.476], Figure 4.10e). At TD, the non-dancers had a significantly larger increase in the backward trunk velocity than the dancers (*p* = 0.002, *d* = 0.985, 95% CI: [0.118, 1.359], Figure 4.10f, Tables A5-A6).

The linear regression model of the faller rate from the first to the last slip revealed that the dancers adapted to the repeated standing-slips faster than the non-dancers (Figure 4.11, Table A7). The slopes of the two linear regression lines (-0.095 for dancers versus -0.012 for non-dancers) differ significantly, with a steeper slope signifying a faster reduction in the fall rate over the repeated standing-slips (*p* = 0.019, Figure 4.11), which indicates a quicker adaptation rate to the slip perturbations among dancers than among non-dancers. In fact, the dancers reduced their faller rate about 8 times faster than non-dancers (-0.095 vs. -0.012, Figure 4.11).

For the percent change in EMG latency from the first to the last slip, the dancers significantly shortened their muscle latencies more compared to non-dancers for the stepping RF (p = 0.004, d = 0.848, 95% CI: [0.217, 1.470], Figure 4.12a) and stepping TA (p = 0.045, d = 0.534, 95% CI: [0.035, 1.267], Figure 4.12c, Tables A8 – A9). No significant group differences were detected for the percent change in muscle latencies for the standing RF (p = 0.139, d = 0.336, 95% CI: [-0.269, 0.938], Figure 4.12a), stepping BF (p = 0.103, d = 0.394, 95% CI: [-0.214, 0.996], Figure 4.12b), standing BF (p = 0.247, d = 0.211, 95% CI: [-0.392, 0.810], Figure 4.12b), standing TA (p = 0.463, d = 0.028, 95% CI: [-0.571, 0.628], Figure 4.12c), stepping MG (p = 0.168, d = 0.298, 95% CI: [-0.307, 0.899], Figure 4.12d), or standing MG (p = 0.260, d = 0.198, 95% CI: [-0.404, 0.798], Figure 4.12d, Tables A8 – A9).

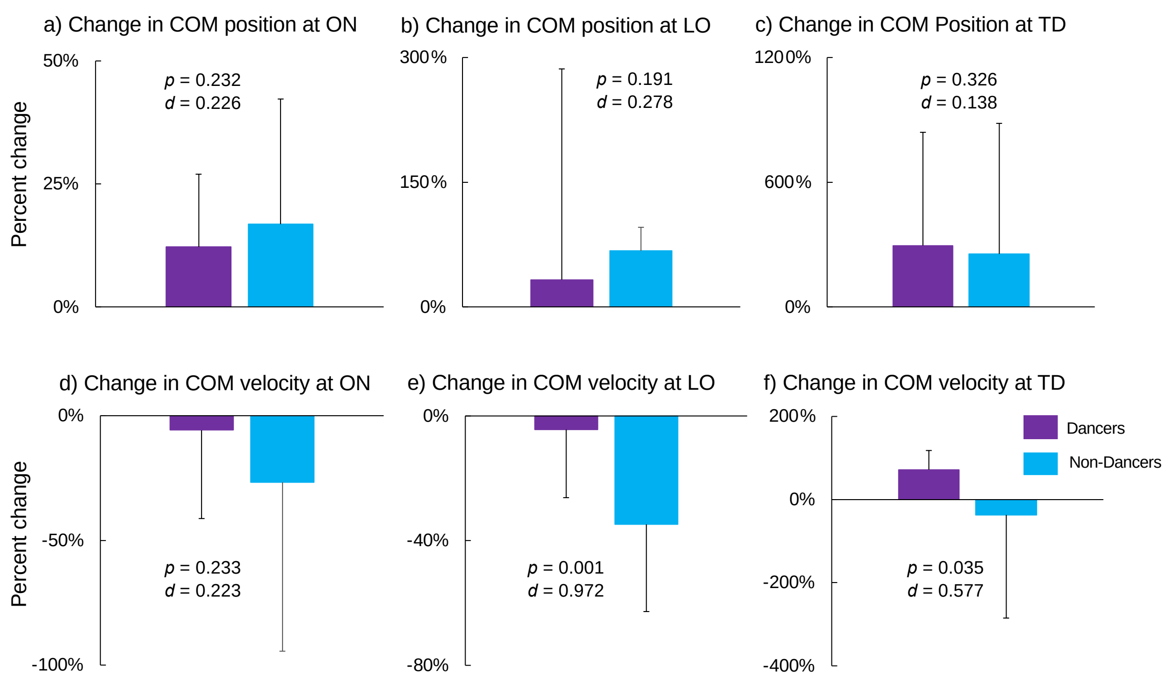

Figure 4.8 Comparisons of the percent change from first to last standing-slip for the center of mass (COM) position at a) slip onset (ON), b) recovery step liftoff (LO), and c) recovery step touchdown (TD), and COM velocity at d) ON, e) LO, and f) and TD between ballet dancers (n = 20) and non-dancers (n = 23). Effect sizes (Cohen's d) are provided. The column height and the error bar represent the group mean and standard deviation, respectively.

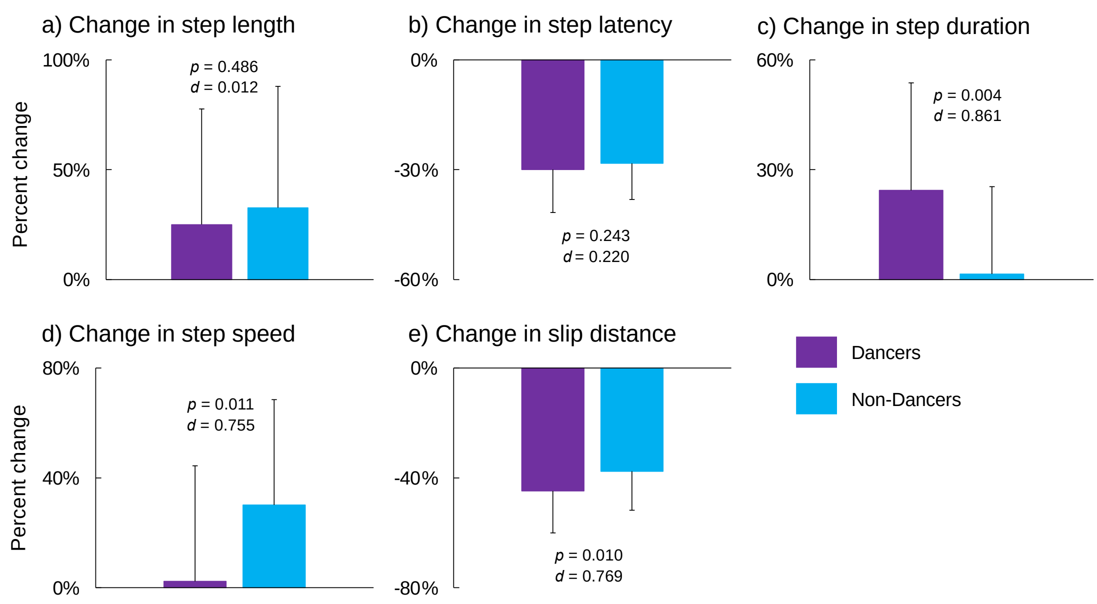

Figure 4.9 Comparisons of the percent change from the first to last standing-slip for the recovery step a) length, b) latency, c) duration, d) and speed, and the e) slip distance between dancers (n = 20) and non-dancers (n = 23). Also shown are the effect sizes in Cohen's d. The column height and the error bar represent the group means and standard deviation, respectively.

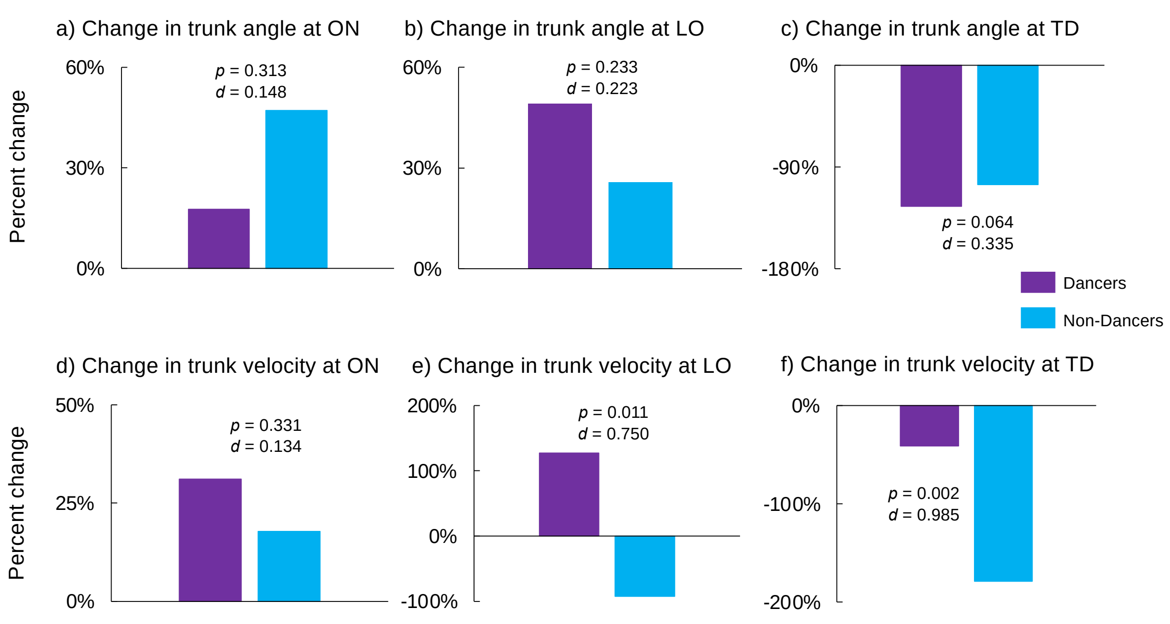

Figure 4.10 Comparisons of percent change from the first to last standing-slip for the trunk angle at the a) slip onset (ON), b) recovery step liftoff (LO), and recovery step touchdown (TD), and the trunk angular velocity at d) ON, e) LO, and f) TD between dancers (n = 20) and non-dancers (n = 23). A negative trunk angle indicates a backward-leaning trunk and a negative trunk angular velocity is in the backward direction. Effect sizes (Cohen's d) are provided. The column height represents the group mean.

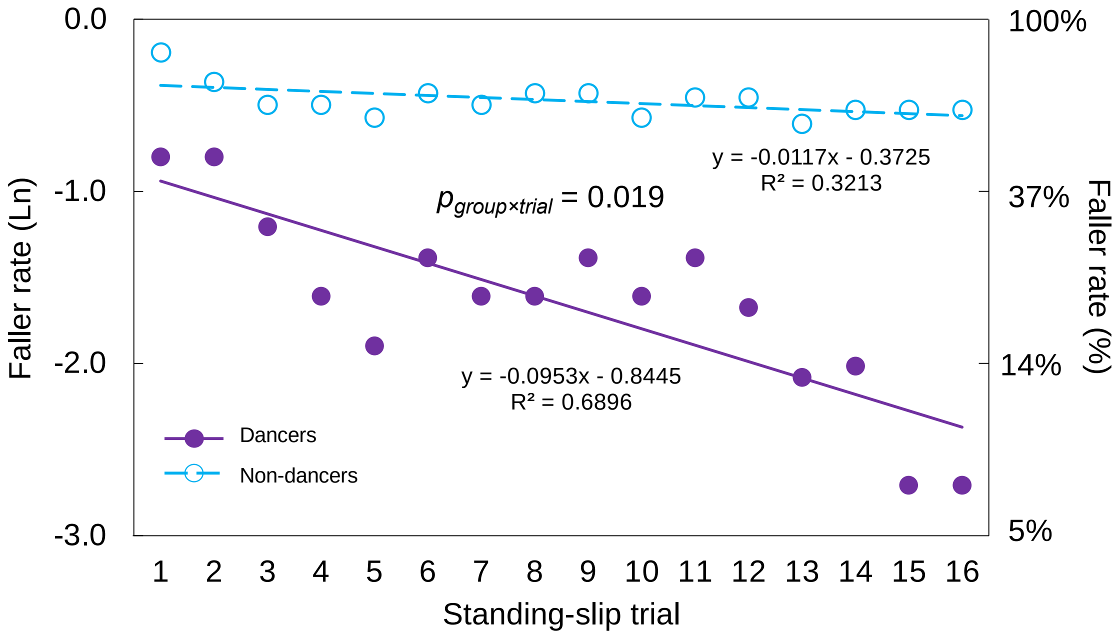

Figure 4.11 Comparisons of the slip outcome adaptation rate from first to last slip between dancers (n = 20) and non-dancers (n = 23). The slip outcome (fall vs. non-fall) was classified as a fall if the peak loadcell force exceeded 30% of the bodyweight. The slip-faller rate was the ratio of the number of fallers to the total participant number in each group. The x-axis shows the standing-slip trial, and the y-axis reflects the faller rate, which was natural log transformed in the left y-axis and the actual values in the right y-axis. The linear regression lines, equations, and R^2 values are displayed for both groups. The faller rate was converted using the natural logarithm function.

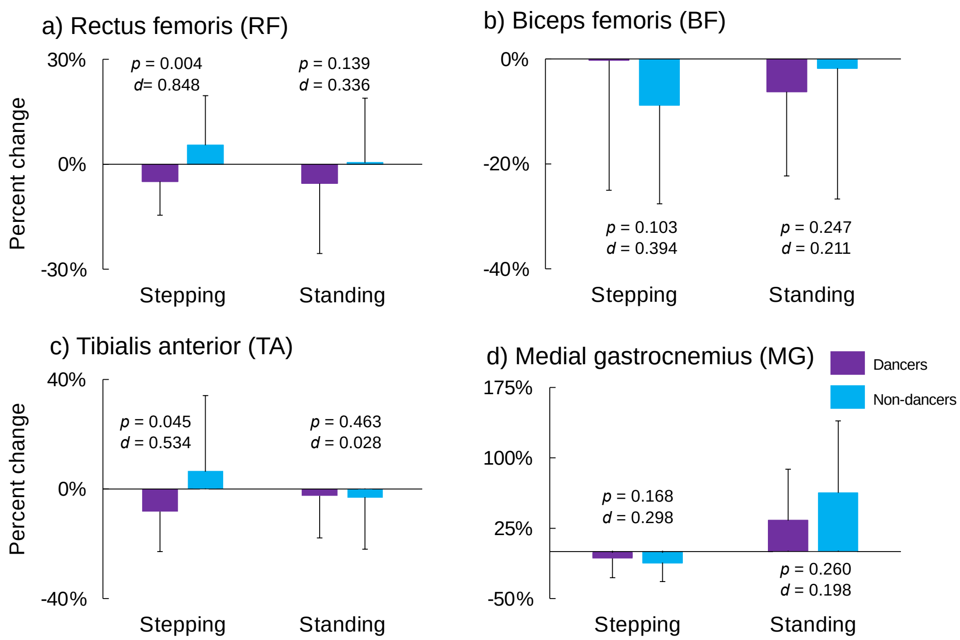

Figure 4.12 Comparisons of the percent change from first to last standing-slip in electromyography (EMG) latencies of the stepping and standing legs of the a) rectus femoris, b) biceps femoris, c) tibialis anterior, and d) medial gastrocnemius between dancers (n = 20) and non-dancers (n = 23). Cohen's d is also presented to quantify the effect size. The column height and the error bar represent the group mean and standard deviation, respectively.

4.7 Hypothesis 3: Transfer (Generalization)

Thirty-six participants (15 dancers and 21 non-dancers) completed the gait-slip trial. Of the other seven participants who completed the standing-slip trials, five (4 dancers and 1 non-dancer) completed their data collection session before the gait-slip was added to the protocol, one participant (1 dancer) requested not to participate in the gait-slip, and one subject (1 non-dancer) was unable to walk comfortably at the treadmill speed necessary (0.8 m/s) for the gait-slip. Thus, the gait-slip trial results represent data from 15 dancers and 21 non-dancers.

Eight variables (stability at RTD, step length, slip distance, trunk angle at RTD and LLO, trunk velocity at LLO, and BF stepping latency) did not meet the normality assumption. Two variables were transformed using the exponential (step length) and the logarithmic base 10 (slip distance) transformations. The remaining six variables could not be transformed to correct normality. These remaining variables were analyzed using the Mann-Whitney *U* test. Three variables (stability at LTD, COM position at LTD, and COM velocity at LTD) violated Levene's test. These three variables were also analyzed with Mann-Whitney *U* tests. Four variables (stability at LLO, COM position at LLO, slip distance, and trunk velocity at RTD) were correlated with body mass. These four variables were analyzed with ANCOVA with body mass as the covariate. No variables were significantly correlated with the total physical activity score. To reduce Type I error due to multiple comparisons, Holm's Sequential Bonferroni correction was made to the significance level for the primary outcome variables (Holm, 1979). The adjusted significance level varied from 0.017 (0.05/3) to 0.05 (0.05/1) for the primary outcome variables. Cohen's *d* and its 95% CI were used to quantify the effect size for all comparisons.

4.7.1 Primary Outcome: Dynamic Gait Stability

For the unexpected gait-slip (GS1), dynamic gait stability was comparable in dancers and non-dancers at LTD ($p = 0.488$, $d = 0.016$, 95% CI: [-0.700, 0.625], Figure 4.13a) and RLO ($p = 0.132$, $d = 0.397$, 95% CI: [0.000, 1.050], Figure 4.13b), but stability was significantly higher in dancers at RTD ($p = 0.012$, $d = 0.814$, 95% CI: [0.261, 1.663], Figure 4.13c, Tables A9-A10), where a higher dynamic gait stability value indicates a more stable state against backward balance falling.

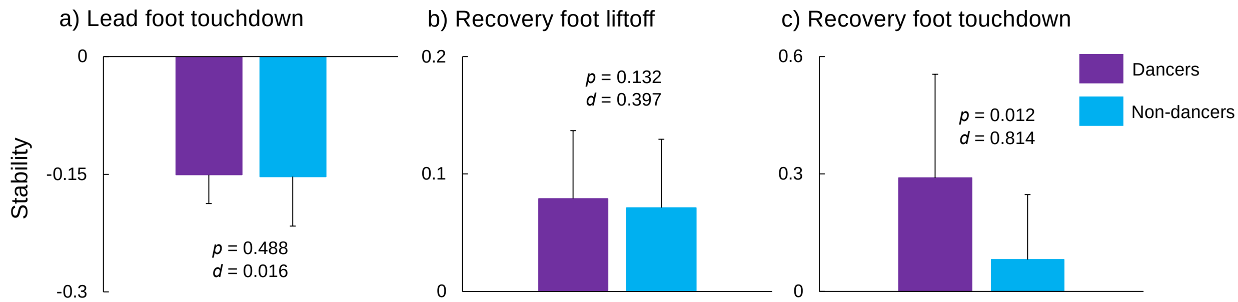

Figure 4.13 Comparisons of the dynamic gait stability at a) lead foot touchdown, b) recovery foot liftoff, and c) recovery foot touchdown between dancers (n = 15) and non-dancers (n = 21). Dynamic gait stability was calculated as the shortest distance from the COM motion state to the threshold against backward balance loss. Effect sizes (Cohen's d) are provided. The column height and the error bar represent the group mean and standard deviation, respectively.

4.7.2 Secondary Outcomes

No significant group differences were identified for COM position at LTD (p = 0.472, d = 0.024, 95% CI: [-0.639, 0.687], Figure 4.14a) or RLO (p = 0.129, d = 0.403, 95% CI: [0.000, 1.054], Figure 4.14b), however, the dancers displayed a significantly more anterior COM position relative to the BOS than the non-dancers at RTD (p = 0.004, d = 0.983, 95% CI: [0.371, 1.793], Figure 4.14c, Tables A9-A10). Both groups had comparable COM velocity at LTD (p = 0.114, d = 0.415, 95% CI: [-0.258, 1.081], Figure 4.14d), RLO (p = 0.488, d = 0.010, 95% CI: [-0.663, 0.663], Figure 4.14e), and RTD (p = 0.223, d = 0.264, 95% CI: [-0.332, 1.003], Figure 4.14f, Tables A9-A10). The dancers displayed a significantly longer (p = 0.002, d = 1.069, 95% CI: [0.352, 1.771], Figure 4.15a) and faster (p = 0.009, d = 0.844, 95% CI: [0.134, 1.516], Figure 4.15c, Tables A9-A10) recovery step than the non-dancers in response to the gait-slip. The recovery step duration (p = 0.219, d = 0.265, 95% CI: [-0.402, 0.929], Figure 4.15b) and slip distance (p = 0.121, d = 0.414, 95% CI: [0.000, 1.070], Figure 4.15d, Tables A9-A10) were similar between groups. The trunk angle was comparable between dancers and non-dancers at LTD (p = 0.306, d = 0.177, 95% CI: [-0.767, 0.559], Figure 4.16a), RLO (p = 0.187, d = 0.310, 95% CI: [-0.797, 0.530], Figure 4.16b), and RTD (p = 0.220, d = 0.265, 95% CI: [-0.403, 0.928], Figure

4.16c, Tables A9-A10). The trunk angular velocity was similar between groups at LTD (p = 0.268, d = 0.220, 95% CI: [0.000, 0.870], Figure 4.16d) and RLO (p = 0.284, d = 0.199, 95% CI: [-0.696, 0.532], Figure 4.16e), however, at RTD the dancers had a significantly slower and less backward trunk velocity than the non-dancers (p = 0.011, d = 0.815, 95% CI: [0.120, 1.500], Figure 4.16f, Tables A9-A10).

In the gait-slip trial, two out of 15 dancers experienced a fall, and nine out of 21 non-dancers experienced a fall. Fisher's exact test compared the gait-slip outcome between groups. The dancers displayed a significantly lower faller rate compared to non-dancers (13.3% vs. 42.9%, p = 0.029, d = 0.565, 95% CI: [-0.362, 1.490], Figure 4.17). For the gait-slip EMG latency, the dancers exhibited significantly shorter muscle latencies than non-dancers in the recovery stepping leg for the RF (p = 0.032, d = 0.717, 95% CI: [-0.042, 1.464], Figure 4.18a), BF (p = 0.031, d = 0.661, 95% CI: [-0.006, 1.512], Figure 4.18b), TA (p = 0.045, d = 0.724, 95% CI: [-0.111, 1.545], Figure 4.17c), and MG (p = 0.038, d = 0.686, 95% CI: [-0.071, 1.431], Figure 4.18d, Tables A11-A12). The EMG latencies for some participants could not be collected because of technical difficulties and equipment malfunction (Figure 4.18).

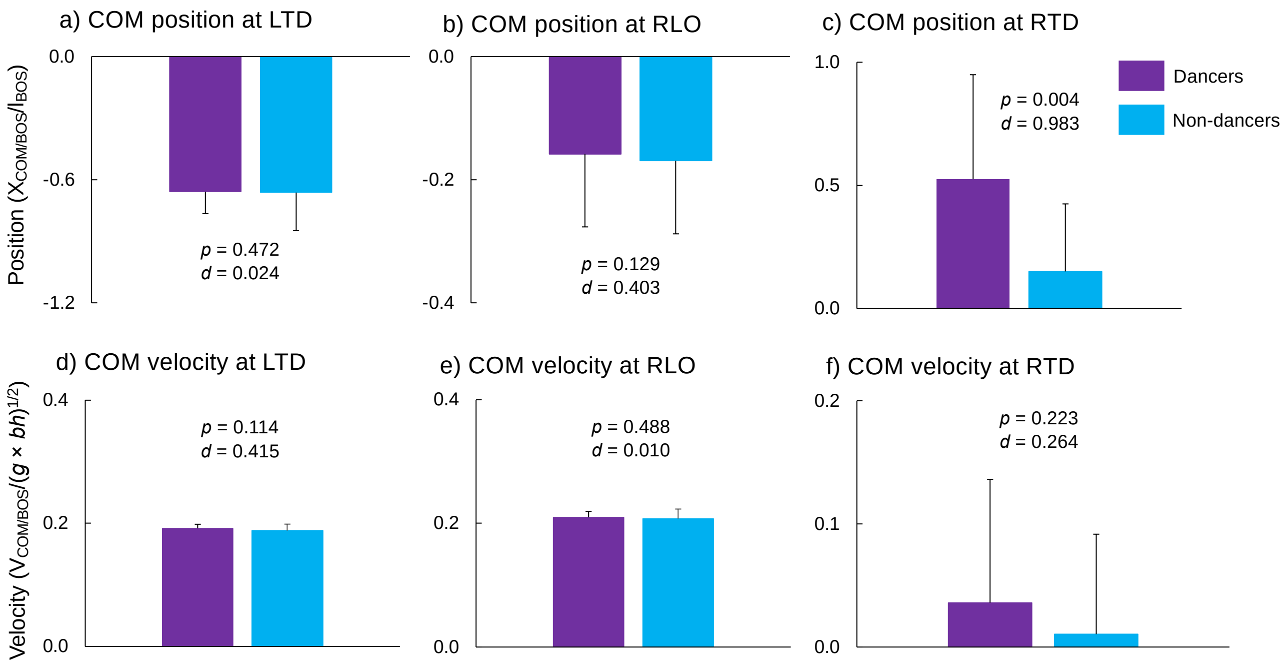

Figure 4.14 Comparisons of the center of mass (COM) position at a) lead foot touchdown (LTD), b) recovery foot liftoff (RLO), and c) recovery foot touchdown (RTD), and the COM velocity at d) LTD, e) RLO, and f) and RTD between ballet dancers (n = 15) and non-dancers (n = 21) upon the gait-slip. The COM position and velocity are relative to the rear of the base of support (BOS) and normalized by the foot length (l_{BOS}) and $\sqrt{g \times bh}$, where g is the gravitational acceleration and bh is the body height. Also displayed are the effect sizes in Cohen's d. The column height and the error bar represent the group mean and standard deviation, respectively.

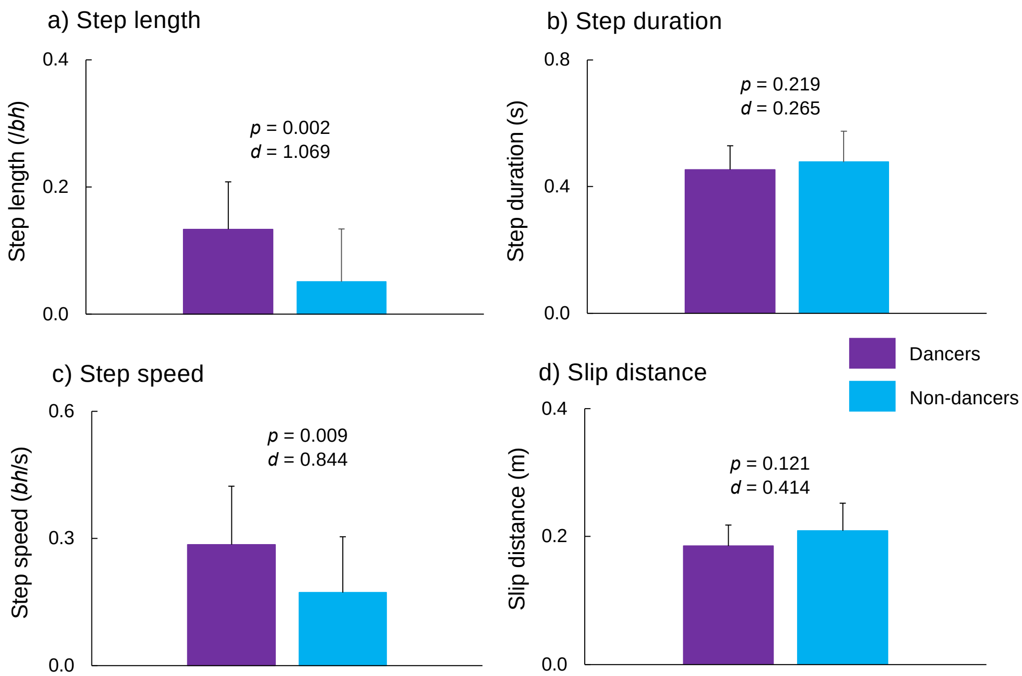

Figure 4.15 Comparisons of the recovery step a) length, b) duration, c) and speed, and d) slip distance between dancers (n = 15) and non-dancers (n = 21) following the gait-slip. Step length was the anteroposterior distance between heels at recovery foot touchdown and expressed in body height (*bh*). Step duration was the interval between recovery foot liftoff and touchdown. Step speed was the recovery step length divided by its duration. Slip distance was the treadmill belt's displacement between slip onset and recovery foot liftoff. Effect sizes (Cohen's *d*) are provided. The column height and the error bar represent the group mean and standard deviation, respectively.

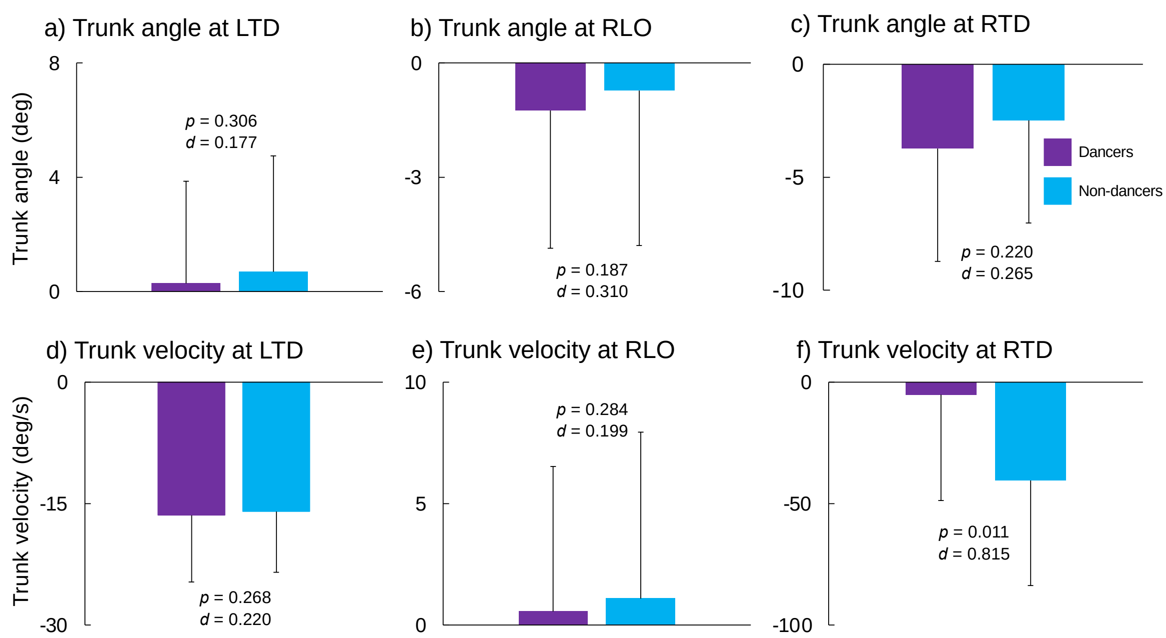

Figure 4.16 Comparisons of trunk angle at a) lead foot touchdown (LTD), b) recovery foot liftoff (RLO), and recovery foot touchdown (RTD), and trunk angular velocity at d) LTD, e) RLO, and f) RTD between dancers (n = 15) and non-dancers (n = 21). The trunk angle in degrees was the angle formed by the trunk segment and the vertical reference line (where a negative angle indicates backward trunk lean). The instantaneous trunk angular velocity (in deg/s) was the first derivative of the trunk angle with respect to time. Effect sizes are provided in Cohen's d. The column height and error bar represent the group mean and standard deviation, respectively.

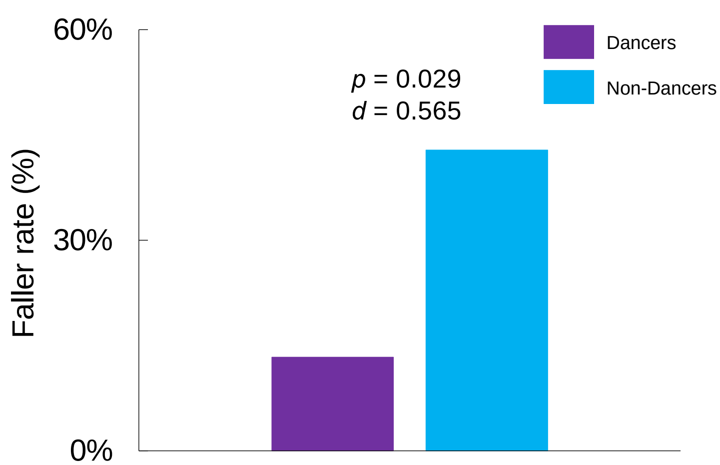

Figure 4.17 Comparison of the gait-slip faller rate between dancers (n = 15) and non-dancers (n = 21) in response to the gait-slip. The slip outcome (fall vs. non-fall) was classified as a fall if the peak loadcell force exceeded 30% of the bodyweight and then was verified by video recordings. The slip-faller rate was the ratio of the number of fallers to the total participant number in each group. The effect size (Cohen's d) is also provided.

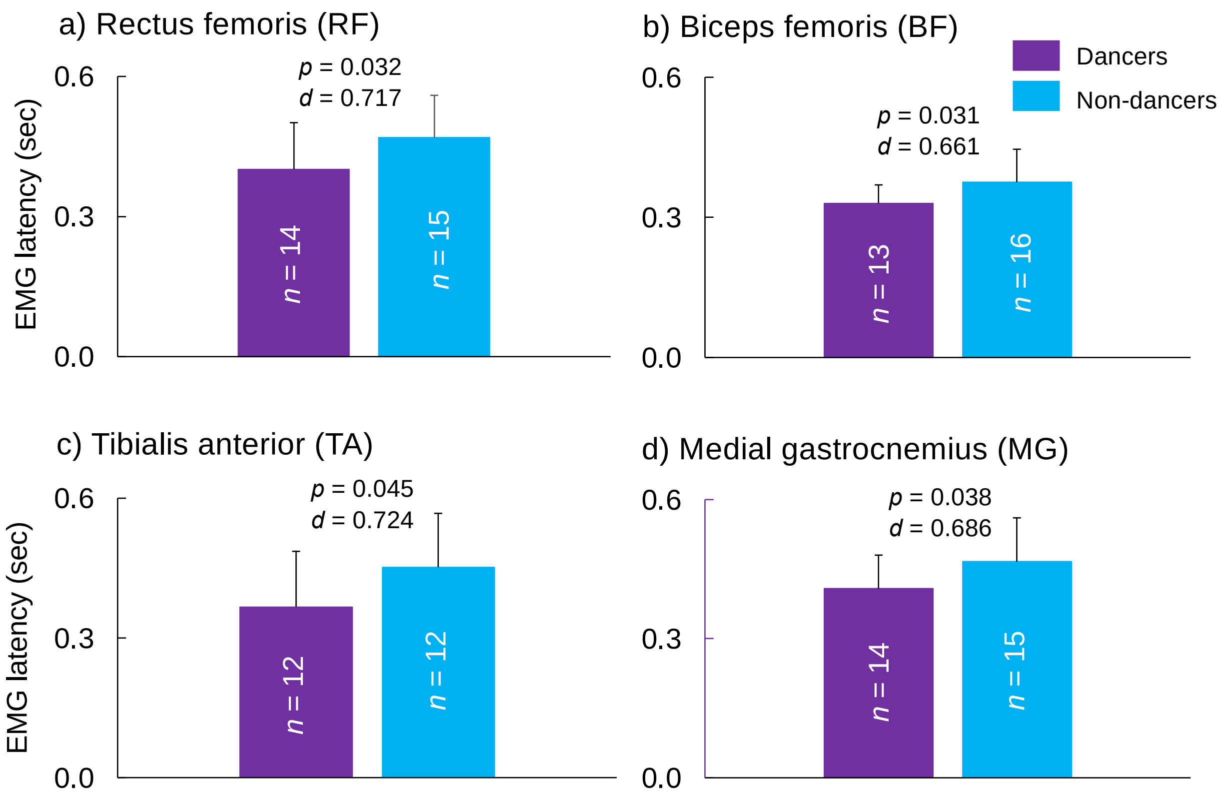

Figure 4.18 Comparisons of electromyography (EMG) latencies during the gait-slip of the stepping leg a) rectus femoris, b) biceps femoris, c) tibialis anterior, and d) medial gastrocnemius between dancers and non-dancers. The EMG latency (sec) for the gait-slip trial was calculated as the time elapsed from the slip onset to the EMG onset, where the EMG onset was determined using a preset threshold and then corrected as needed following visual inspection. The number of usable participants (*n*) and effect sizes (Cohen's *d*) are provided. The column height and the error bar represent the group mean and standard deviation, respectively.

5 DISCUSSION

The overall purpose of this study was to examine the association between ballet practice and improvements in the reactions to large-scale external slip perturbations as related to slip-fall risk among older adults from a neurobiomechanical perspective. The objective was threefold:

1. To examine how older ballet dancers respond to the first well-controlled, novel, and large-scale slip perturbation during standing compared to non-dancers.
2. To determine whether older ballet dancers adapt to repeated standing-slip perturbations more quickly than their non-dancer counterparts.
3. To investigate if older ballet dancers can transfer the fall resistant skills acquired from the repeated standing-slips more effectively to a novel gait-slip than non-dancers.

Correspondingly, the tested hypotheses included:

1. Ballet dancers would experience fewer falls with higher stability, a more effective recovery step, better controlled trunk movement, and shorter leg muscle EMG latencies after the first unexpected standing-slip compared to non-dancers.
2. Ballet dancers would show a quicker adaptation to repeated standing-slip perturbations than non-dancer controls. Specifically, dancers would exhibit greater improvement in their slip-reactions (including slip-faller rate, dynamic gait stability, trunk movement, and EMG latency) compared to the non-dancers.
3. Ballet dancers would show more transfer of the fall resistant skills learned from standing-slips to the novel gait-slip than their non-dancer counterparts. In detail, dancers would experience fewer falls with higher stability, a more effective recovery step, better controlled trunk movement, and shorter leg muscle EMG latencies after the novel gait-slip compared to non-dancers.

5.1 Hypothesis 1: Reaction

The results partially supported the first hypothesis. The dancers experienced significantly fewer falls than the non-dancers (45% vs 83%) in response to the first unexpected standing-slip. Based on this finding, the odds of an older ballet dancer falling after an unexpected standing-slip perturbation are lower than that of a non-dancer (odds ratio or OR = 0.17). This dancers versus non-dancers OR is lower than the one a recent meta-analysis reported (OR = 0.64) to demonstrate the effects of regular traditional exercise relative to controls in reducing falls (Papalia et al., 2020). Such a comparison implies that ballet practice may lead to a better outcome in reducing fall risk in older adults than regular traditional exercises.

The difference in faller rate in the current study could be explained by the group differences in the primary outcome measure of dynamic gait stability. The ballet dancers exhibited significantly higher stability than their non-dancer counterparts at LO and TD. According to the FSR theoretical framework (section 2.6), a higher value of dynamic gait stability indicates a more stable state against backward falling. The dancers and the non-dancers displayed comparable positive stability at ON, and this finding aligns with a previous study reporting similar stability at ON before an unexpected standing-slip in young adult professional ballet dancers compared to non-dancers (Simpkins et al., 2022a). Alternatively, a negative stability value represents a COM motion state below the FSR, and a lack of sufficient forward momentum in the body to shift the COM above the BOS (Yang et al., 2007). Both groups displayed negative stability values at LO and TD (Table A2). Therefore, both groups experienced backward balance loss and implemented a backward recovery step in response to the first unexpected standing-slip. However, dancers were significantly more stable with medium to large effect sizes than non-dancers at both instants (LO and TD) of the first recovery step given that their stability values were closer to the FSR lower boundary (less negative) than the non-dancers.

The COM motion state used to determine stability is comprised of two components (COM position and velocity), therefore the examination of group differences in these two outcomes is meaningful to further explain the observed differences in stability. The COM position was significantly more anterior (positive) in dancers than non-dancers at ON, LO, and TD. According to the FSR, a value of zero on the *x*-axis indicates a COM position located directly above the heel, a value of one indicates a COM position located directly above the toes, and a value of negative one indicates a COM position located one foot's length behind the heel (Figure 2.1). For LO and TD, both groups exhibited negative COM positions falling outside of (and behind) the BOS. However, the COM position value for the dancers was less negative (or closer to the FSR lower boundary) than that of the non-dancers. The more anteriorly located COM positions (related to the BOS) of the dancers contributed to their greater stability at both LO and TD. For COM velocity, both groups had comparable COM velocities at ON and TD (and with very small to small effect sizes), but at LO the dancers exhibited a faster backward COM velocity than the non-dancers. Since the slip perturbations in this study prompted balance loss in the backward direction, a faster backward COM velocity could potentially lead to greater instability (a COM motion state farther away from the FSR lower boundary) if the accompanying COM position is not located anteriorly enough to counteract it. Thus, although the dancers' COM velocity at LO was significantly more negative than the non-dancers, their more positive COM position at LO resulted in a higher stability value than that of the non-dancers.

The ability to react to a perturbation with a quick and effective step is imperative to avoiding a fall (Cham and Redfern, 2001), and one factor leading to the higher stability at LO among the dancers could be their more successful recovery step initiation as evidenced by their shorter step latency than the non-dancers. A promptly initiated recovery step has the potential to place the COM closer to the BOS at LO, thus improving the body's resilience against slip-related backward balance loss. The dancers initiated their recovery step significantly sooner than the non-dancers after ON, causing them to have better stability and more successfully counteract the slip perturbation. The shorter step latency of the dancer group also contributed to their faster backward COM velocity at LO. More specifically, the standing-slip profile for the treadmill was designed to reach a peak forward treadmill belt velocity of 1.2 m/s at 0.3 seconds and then decelerate the belt back to 0 m/s in the following 0.3 seconds (Figure 3.3b). Both the dancers and non-dancers initiated their recovery step after the treadmill belt peak velocity at 0.3 seconds (as evidenced by their step latencies of 0.344 ± 0.053 seconds and 0.378 ± 0.062 seconds, respectively), however, the dancers initiated their step sooner after this peak treadmill velocity than the non-dancers. During the decelerating phase of the treadmill belt movement, the later the time, the slower the belt speed. Therefore, the BOS velocity at the LO was larger for dancers than non-dancers. Despite the faster backward moving COM relative to the BOS among dancers than non-dancers, which potentially impairs their dynamic gait stability, dancers exhibited a significantly more anteriorly located COM relative to the BOS than the non-dancers, making the dancers more stable at this instant than their non-dancer peers.

The dancers displayed a significantly shorter step duration and a faster step speed than the non-dancers during the first recovery step, which contributed to their higher stability at TD. This more efficient reactive response among the dancers could be related to their ballet practice. Ballet involves the repetition of many movements in the backward direction (Kostrovitskaya and Pisarev, 1978), and individuals with ballet experience could be more confident in taking a faster backward step during the backward balance loss recovery process. Faster backward recovery steps were also previously reported in young professional ballet dancers compared to young non-dancers following an unexpected standing-slip (Simpkins et al., 2022a). Since typical human locomotion involves forward stepping, the older non-dancers in the current study were possibly less experienced in stepping backwards, leading to a later initiated step and a slower step than the dancers. This lack of backward-based movement experience could potentially predispose non-dancers to a higher risk of experiencing a backward fall when exposed to a slip during standing.

The recovery step length was comparable between groups with a very small effect size following the first standing-slip trial. This finding differs from a previous study that investigated an unexpected standing-slip perturbation response in young professional ballet dancers and reported a significantly longer backward recovery step in dancers versus non-dancers (Simpkins et al., 2022a). This discrepancy could be due to two reasons. First, the experience levels of ballet dancers differ greatly between the previous study (professional dancers) and the current study (mainly recreational dancers). Ballet positions with the “gesturing limb” held posteriorly are more unstable than other limb orientations (anterior or lateral), and previous work reported that more experienced dancers are more efficient at controlling balance in such positions than less experienced dancers (Bruyneel et al., 2010). Therefore, it is possible that the older dancers in the current study took a backward recovery step of a similar length to their non-dancer counterparts given their more recreationally focused experience with ballet. Second, the age groups of the previous study (young adults) and the current one (older adults) also differ. Given that aging is associated with shortened step lengths (Laufer, 2005), and previous work has reported shorter step lengths in older adults than young adults when exposed to an unexpected slippery walking surface (Moyer et al., 2006), it is also possible that the observed step length similarity between the dancers and non-dancers in the current study was due to both groups being at a more advanced age with comparable step length shortening.

Another factor that can assist with successful balance restoration following a slip perturbation is effective trunk movement control, and trunk movement is associated with dynamic gait stability. This is because the head-arm-trunk segment contains roughly 2/3 of the body mass and has a substantial impact on the COM motion state (Yang and Pai, 2013). Furthermore, the ability to terminate the excessive backward trunk rotation following a slip can play an important role in avoiding a fall (Grabiner et al., 2008). In the current study, no significant group differences were found for trunk angle and angular velocity at ON, LO, or TD. Thus, older adult ballet dancers did not display significantly better trunk control than non-dancers in response to an unexpected standing-slip. This finding also conflicts with the aforementioned previous study that exposed young professional ballet dancers to an unexpected standing-slip and reported significantly more upright trunk angles at ON, LO, and TD, and a slower backward trunk velocity at TD in the dancers (Simpkins et al., 2022a). However, again, this discrepancy could be due to differences in ballet experience and age groups between the previous study (professional young adults) and the current one (recreational older adults). Of note, though, the dancers were more upright with less backward-leaning trunks than the non-dancers at all three instants in the current study, although not to a statistically significant extent. The dancers' more erect trunk position likely contributed in part to their more anterior COM positions for ON, LO, and TD, which ultimately led to better stability values, as well. Additionally, the more upright trunk position of the dancers could be related to their ballet training, which emphasizes an erect posture while maintaining the body weight mostly over the balls of the feet, allowing for ease of movement, efficient directional changes, and proper lower extremity alignment (Clippinger, 2007).

EMG analysis has been widely used to investigate both successful (Chambers and Cham, 2007; Marigold and Patla, 2002) and unsuccessful (Qu et al., 2012) balance recovery following a slip (Ahn et al., 2024b). In the current study, the latencies of four lower extremity muscles (RF, BF, TA, and MG) in both legs were examined, as these four muscles are the major muscle groups controlling lower extremity movement. The latency of the leg muscles could be explained from the aspect of their anatomic structures. For the recovery stepping leg, the muscle latencies of the BF, TA, and MG were significantly shorter in dancers than in non-dancers. The BF serves as a knee flexor and a hip extensor, and the TA is an ankle dorsiflexor (Qu et al., 2012). The dancers activated the BF and TA faster upon detecting the slip and began bending the knee, extending the hip, and flexing the foot to lift the foot and clear the floor sooner than the non-dancers during their backward recovery step. The flexed knee and dorsiflexed ankle joints would also reduce the moment of inertia of the leg around the hip joint, facilitating the initiation and execution of the recovery step (Ahn et al., 2024a). While a primary action of the MG is foot plantarflexion, this muscle also contributes to knee flexion (Qu et al., 2012). The shorter stepping leg MG latency in the dancers than the non-dancers potentially further assisted with knee flexion of the recovery stepping leg to bend the knee and clear the floor. However, the findings for the stepping MG should be interpreted with caution. The stepping MG data were only successfully collected for 14 (out of 20) dancers and 15 (out of 23) non-dancers in the first standing-slip trial due to equipment malfunction and technical errors. These stepping MG findings should be further verified in future studies with large sample sizes. No group differences were found for the latency of the stepping RF. This RF finding is reasonable for the stepping leg since knee extension would be undesirable for the initiation of the backward recovery step.

In the standing leg, the dancers displayed significantly shorter RF and TA latencies compared to the non-dancers. In the standing leg, a more quickly activated RF could help to maintain knee extension to provide support and prevent standing limb collapse following the standing-slip. With respect to the TA, a shorter muscle latency in the standing TA could assist with stabilizing the standing ankle joint in response to the slip perturbation. The standing TA could additionally act as a primary agonist muscle that attempts to slow down the backwardly rotating shank after the forward slip perturbation (Welch and Ting, 2009). Successful balance recovery following a slip has been previously associated with increased ankle muscle co-contraction (Chambers and Cham, 2007; Qu et al., 2012). While no significant group difference and very small effect size were found for the standing leg MG latency, the dancers activated their standing MG relatively sooner than the non-dancers, perhaps creating a TA-MG co-contraction to improve ankle stabilization. The stabilized standing leg would provide a stationary base for participants to successfully perform their recovery stepping.

Due to EMG technical difficulties and equipment malfunction, EMG data for the standing MG were only successfully collected for 11 out of 20 dancers. Thus, a significant between-group difference could emerge with a larger sample size. However, this notion must be investigated in future projects. Finally, no significant group difference was discovered for the standing leg BF. Given that the BF contributes to knee flexion and hip extension, a shortened standing BF latency would not be advantageous in the case of a standing-slip balance loss.

Overall, the results for the first standing-slip trial suggest that older ballet dancers can better control the body's response to an unexpected slip perturbation during standing, which reduces their fall risk (Figure 5.1). For dancers, their higher stability at LO and TD could be attributed to faster and more effective recovery stepping. The more effective recovery stepping may be associated with the earlier activated leg muscles in dancers than in non-dancers. These better reactions to the novel standing-slip in dancers could be associated with their ballet experience which involves frequent backward stepping.

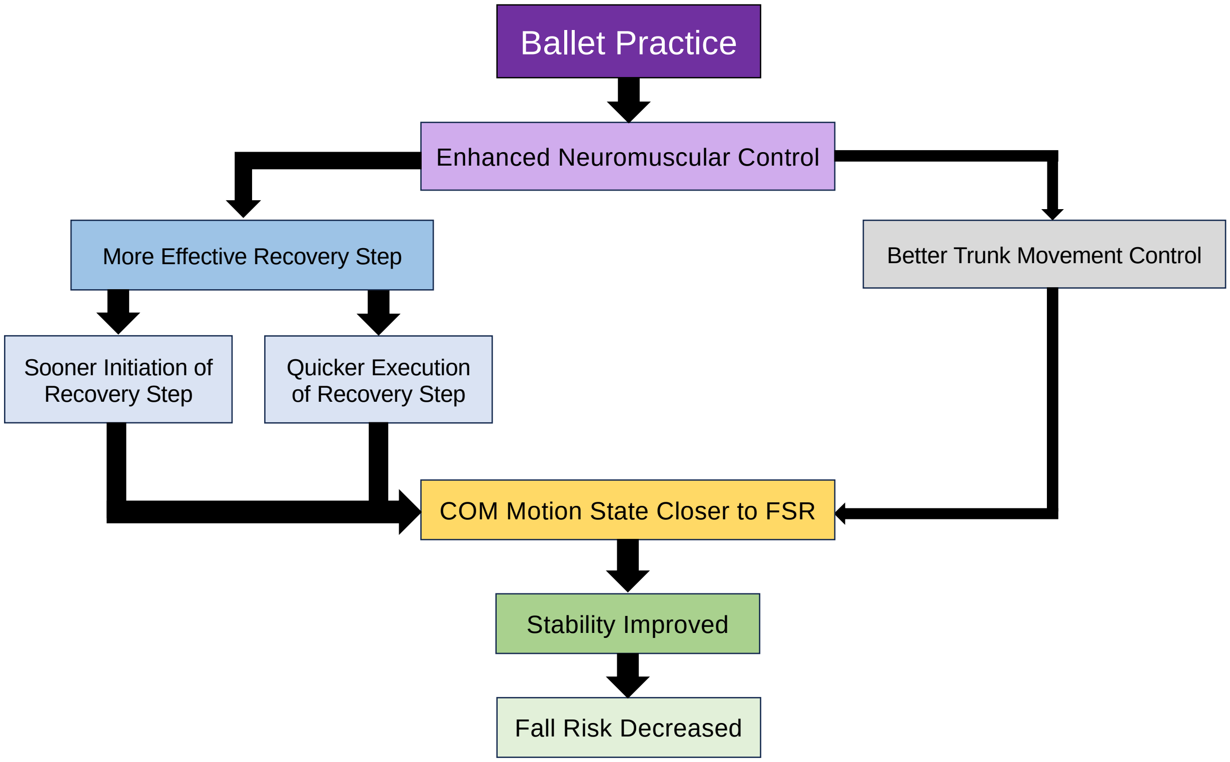

Figure 5.1 Pathways showing the proposed relationship between ballet practice and reduced fall risk for older adults from a neurobiomechanical perspective.

5.2 Hypothesis 2: Adaptation

The results partially supported the second hypothesis. From the first to the last standing-slip, the dancers' faller rate was reduced by 38% (from 45% to 7%), and the non-dancers' faller rate was reduced by 24% (from 83% to 59%). As evidenced by the linear regression analysis, the dancers adapted faster to the repeated standing-slips and improved their faller rate by the final slip more than the non-dancers.

The primary outcome results indicated that dancers displayed a more effective reactive adaptation than non-dancers, but a similar proactive adaptation in response to the repeated standing-slip trials. From the first to last standing-slip trial, the dancers significantly improved their dynamic gait stability at LO more than the non-dancers. Given that the recovery stepping event of LO occurs after the onset of the slip, this instant can be used to characterize the reactive control of dynamic balance. In contrast, there was no group difference in the adaptation from the first to last standing-slip for stability at ON (very small effect size) or TD (small effect size). Both groups increased their stability at ON (becoming more stable) and therefore used similar proactive strategies to adapt to the repeated slip perturbations. Both groups also increased their stability at TD from the first to last standing-slip to a statistically similar extent. These findings for the primary outcome measures concur with a previous study that reported better reactive control but similar proactive control in young professional ballet dancers following repeated standing-slips (Simpkins and Yang, 2023a).

No group differences were detected in percent change for COM position at ON, LO, or TD. Both groups comparably increased their COM position at all three instants, indicating a more favorably (anteriorly) positioned COM at the final slip compared to the first slip. There was no group difference in the percent change of COM velocity at ON, however, significant group differences emerged for the percent change of COM velocity at LO and TD. At LO, the dancers maintained a similar COM velocity from first to last slip (decreased and became more negative by ~4%), whereas the non-dancers decreased their COM velocity to a greater extent (decreased and became more negative by ~35%). Given that a more negative COM velocity indicates a faster backward-moving COM, a large decrease in COM velocity is not favorable for backward balance loss. At TD, the dancers increased their COM velocity from first to last slip (increased and became less negative by ~72%) and the non-dancers decreased their COM velocity (decreased and became more negative by ~37%). Again, this finding is more favorable for the dancers since a COM moving backward quickly relative to the BOS could be detrimental to recovering balance following a slip perturbation.

Regarding the recovery stepping variables, the dancers significantly increased their step duration more than the non-dancers from the first to last standing-slip with a large effect size. Ballet involves the frequent practice of movements in the backward direction as well as movements standing on one leg (Kostrovitskaya and Pisarev, 1978). Therefore, people with ballet experience could be more confident and comfortable with taking a backward recovery step of a longer duration (spending a longer time standing on one leg) after exposure to repeated standing-slips. The non-dancers in the current study were possibly less experienced with backward stepping since typical human locomotion involves stepping in the forward direction. Additionally, the non-dancers significantly increased their step speed from first to last standing-slip (increased by ~30%) compared to the dancers (increased by ~2%). The dancers took a faster recovery step than the non-dancers in response to the novel unexpected standing-slip, indicating a more efficient reactive response of the dancers to the first standing-slip trial. Given that the dancers already took a fast recovery step before repeated standing-slip exposure, it appears that further increasing their step speed was not necessary for their adaptation response. Alternatively, since the non-dancers took slower recovery steps at the first standing-slip, they adapted to the multiple slip exposures by increasing their backward stepping speed from the first to the final slip.

The dancers also significantly shortened their slip distance more than the non-dancers from the first to last standing-slip. The reduced slip distance made the COM closer to the BOS at LO, thus improving stability and putting the dancers in a more advantageous situation against backward balance loss than the non-dancers after the slip. This finding coincides with previous work on adaptation to multiple standing-slips in young professional ballet dancers which reported significantly decreased slip distances in dancers compared to their young adult counterparts (Simpkins and Yang, 2023a). Similar findings regarding shortened slip distances have also been reported in other work on reactions to repeated slips during gait or sit-to-stand in both healthy and clinical adults (McCrum et al., 2017; Wang et al., 2011; Yang et al., 2019). Finally, while both groups increased their step length and shortened their step latency from the first to the last standing-slip, they did so to a similar extent. Thus, the adaptations in step length and step latency following repeated standing-slips are comparable between healthy older dancers and non-dancers.

There were no significant differences between the change in trunk angles of the dancers and non-dancers at ON, LO, or TD. These results suggest that the trunk orientation did not significantly affect the adaptation rate of either group in response to the repeated standing-slips. However, the changes in trunk angle velocity were significantly different between groups at LO and TD. At LO, the dancers significantly increased their trunk angular velocity in the forward direction more than the non-dancers from first to last standing-slip. This assisted the dancers during their slip recovery since the slip perturbation prompted a balance loss in the backward direction. The change in trunk angular velocity from first to last standing-slip was also significantly different between groups at TD. While both groups increased their trunk velocities in the backward direction, the non-dancers had a significantly larger increase in backward trunk velocity at TD than the dancer group. This increase in backward trunk velocity during a backward balance loss could be highly detrimental to the slip recovery process, and this observation in the non-dancers could explain their much higher faller rate at the last standing-slip compared to the dancers (59% in non-dancers versus 7% in dancers).

Four lower extremity muscles were examined bilaterally in the current study: RF, BF, TA, and MG. Two muscle latencies changed in significantly different ways between groups from the first to the last standing-slip. For the stepping RF and the stepping TA, the latencies were shortened for the dancers and increased for the non-dancers. Quicker activation of the stepping RF and TA could assist in hip flexion and ankle dorsiflexion, respectively, to lift the stepping foot/leg off the treadmill belt and successfully clear the floor during the backward recovery step. Previous work reported shorter muscle latencies (including the RF and TA) for successful balance recoveries in young adults following a gait-slip (Qu et al., 2012). Though not statistically significant, the dancers also shortened the standing RF latency more than the non-dancers from the first to last standing-slip, which potentially aided in the standing leg limb support of the dancers via knee extension. Both groups shortened the latencies of the stepping and standing BF, the standing TA, and the stepping MG to a similar extent. For the standing MG, both dancers and non-dancers lengthened the muscle latency from the first to the final standing-slip. Since the standing TA latencies shortened for both groups and the standing MG latencies lengthened for both groups (although not significantly for either), this could be explained by the fact that during the standing-slip adaptation process, both dancers and non-dancers relied less on the standing leg TA-MG co-contraction to stabilize the standing ankle joint during slip recovery.

In summary, the faller rate of the dancers reduced more than the non-dancers from the first to the last standing-slip trial, signifying a faster adaptation rate for dancers than the non-dancers. The larger improvement in dynamic gait stability at LO in the dancers compared to the non-dancers revealed a more effective reactive adaptation in dancers versus non-dancers, but a similar proactive adaptation to repeated slips during treadmill standing. The dancers increased their step duration and step speed more than the non-dancers across the multiple standing-slip trials, and dancers displayed better improvement in trunk control at LO and TD. Additionally, the dancers shortened the muscle latencies for the stepping TA and the stepping RF more than the non-dancers from the initial to final standing-slip, which contributed to their improved faller rate and more effective recovery stepping.

5.3 Hypothesis 3: Transfer (Generalization)

The results partially support the third hypothesis. The dancers experienced significantly fewer falls following the gait-slip trial (13% vs 43%, $p = 0.029$). This difference in faller rate could be associated with the higher dynamic gait stability at RTD in the dancers compared to their non-dancer counterparts. A higher value of stability in the FSR framework indicates a more stable state against backward balance loss. While both groups displayed a positive stability value at RTD, the dancers' stability was significantly more positive, indicating a more stable state against backward balance loss at the completion of the recovery step. This observation could be interpreted as the dancers transferring their standing-slip training to a novel-gait slip more effectively than the non-dancers for RTD. Previous work has also reported better transfer and stability at RTD in young adults who underwent sit-to-stand perturbation training and then experienced a gait-slip (Wang et al., 2011). Similar stability and very small to small effect sizes were found between dancers and non-dancers at LTD and RLO in the current study. Thus, the standing-slip training did not seem to have a differing transfer effect on dancers versus non-dancers for these two moments.

The dancers' higher stability at RTD is due to their significantly more anterior COM position at the same instant. The COM motion state that determines dynamic gait stability is composed of both the COM position and velocity. Given that the COM velocity at RTD was comparable between groups and the effect size was small, the better stability of the dancers at RTD is due specifically to a more forward COM position. While both groups displayed a COM position value between 0 and 1 at RTD, the dancers had a significantly higher value falling more within the BOS compared to the non-dancers. Alternatively, the COM position and velocity were comparable between dancers and non-dancers at LTD and RLO, which aligns with the finding that the stability was also comparable at these moments.

Another contributing factor to the higher stability at RTD among dancers could be their more successful recovery step than the non-dancers. In response to the gait-slip, the dancers displayed a significantly longer step length and a faster step speed than the non-dancers. By executing a longer and faster step, the dancers enlarged their BOS more effectively than non-dancers, bringing their COM motion state more inside the FSR and improving stability (Espy et al., 2010). This more efficient reactive response among the dancers could be a result of their ballet practice that involves many movements in the backward direction (Kostrovitskaya and Pisarev, 1978). Ballet training experience could make older adult dancers more comfortable and confident in taking a larger and faster backward step when recovering from a backward balance loss. Aligning with this theory, previous work reported an increased backward step length after dance training in people with Parkinson’s disease (Hackney and Earhart, 2009). Again, normal human locomotion is associated with forward stepping. The non-dancers were possibly less comfortable in taking a backward step or moving in a backward direction and therefore took a smaller and slower step, which may not be sufficient to restore balance. A smaller and slower step also predisposes them to a higher backward fall risk. The step duration and slip distance were similar between dancers and non-dancers following the gait-slip, and both groups seemed to transfer their standing-slip training comparably in terms of these stepping metrics.

As mentioned previously, the ability to terminate the excessive backward trunk rotation following a slip can play an important role in avoiding a fall (Grabiner et al., 2008), and trunk movement is associated with dynamic gait stability. Thus, the control of the trunk is another factor that can help with successful balance restoration following a slip perturbation. Similar to the findings for the novel standing-slip trial, the trunk angle was comparable between dancers and non-dancers with very small to small effect sizes at all three moments of interest for the gait-slip (LTD, RLO, and RTD). Additionally, the trunk angular velocity was similar between groups at LTD and RLO. However, the dancers displayed significantly slower and less backward trunk angular velocity at RTD compared to their non-dancer counterparts. A slower backward trunk velocity during a backward balance loss is preferable for the slip recovery process. This observation related to trunk control at RTD in the dancers could help further explain their much lower faller rate compared to the non-dancers (13% in dancers versus 43% in non-dancers). Furthermore, this finding aligns with a previous study that exposed young professional ballet dancers to a novel standing-slip and reported a marginally slower backward trunk velocity at the TD of the recovery foot in the dancers (Simpkins et al., 2022a).

The muscle latencies of four lower extremity muscles (RF, BF, TA, and MG) of the recovery leg were examined during the gait-slip. All four muscle latencies were significantly shorter in dancers than their non-dancer peers. The gait-slip perturbation in the current study was delivered in a standardized manner to all participants at the early- to mid-stance phase during normal treadmill walking. Accordingly, participants were standing completely on one leg with the other leg (the "stepping leg") starting to swing through when they experienced the slip. The first muscle to be activated in both groups after the gait-slip onset was the BF (a knee flexor), followed by the TA (an ankle dorsiflexor). The shorter BF and TA latencies of the dancers allowed them to flex their knee and dorsiflex the ankle on the recovery leg sooner than the non-dancers to move their stepping leg backward and clear the floor to begin their recovery step. As stated, the dorsiflexed ankle and flexed knee reduce the moment of inertia of the leg around the hip joint, which facilitates the initiation and execution of the recovery stepping. The BF and

TA activations were then followed by the RF and MG, which also had significantly shorter latencies in the dancers compared to the non-dancers. It is likely that RF and MG activation at the end of the recovery step extends the knee and plantarflexes the ankle to re-establish contact of the recovery foot with the treadmill belt in order to complete the recovery step. These shorter muscle latencies of the dancers may have contributed to their significantly lower faller rate than the non-dancers (43% vs 13%).

These EMG latency findings are in line with a previous study that reported shorter muscle latencies of the stepping leg (RF, BF, TA, and MG) in successful balance recoveries compared to unsuccessful ones following a slip during walking (Qu et al., 2012). Although Qu et al.'s study design (overground gait-slips in young adults) differs from the current study's design (treadmill gait-slips in older adults), it could be that this muscle activation pattern holds true for various age groups and slipping conditions who successfully recover from a slip during walking. However, the EMG latency findings for the gait-slip in the current study should be interpreted with caution. Due to equipment malfunction and technical errors, EMG latency data were not successfully collected for all muscles in all participants who completed the gait-slip trial. Thus, these gait-slip EMG latency findings of the recovery stepping leg must be further verified in future studies with large sample sizes.

One potential limitation related specifically to this study's third aim is that both the dancer group and the non-dancer group underwent the repeated standing-slip procedure before exposure to the novel gait-slip. The better responses to the novel gait-slip among the dancers than among their non-dancer peers could result from at least two sources: the better transfer of fall resistant skills from the standing-slip training to the gait-slip test and an inherently more effective response to the novel gait-slip. The current study design could not separate these two sources and may not fully capture the motor skills transfer of older adult ballet dancers relative to the non-dancers. Future work should strive to understand the relationship more thoroughly between ballet practice in older adults and the transfer of standing-slip training to a novel gait-slip. Ideally, the study design would include two groups of older adult ballet dancers and two groups of older non-dancers. One group of dancers and one group of non-dancers would undergo the standing-slip procedure followed by a novel gait-slip. The remaining two groups would not experience any standing-slip but only the same gait-slip as the other groups. The comparison of the reactions to the gait-slip among these four groups by a 2 × 2 ANOVA (two factors: dancer condition and training condition) or similar statistical approaches would sufficiently answer the question of whether ballet practice could facilitate fall avoidance skills transfer between contexts in older adults.

In summary, the dancers experienced fewer falls than the non-dancers following the gait-slip. This difference in faller rate could be associated with the dancers' better stability at RTD, their longer and faster recovery steps, and their slower and less backward trunk angular velocity at RTD. The dancers also displayed significantly shorter muscle latencies for all four muscles of the recovery stepping leg compared to the non-dancers. Altogether, these findings show more transfer of anti-fall skills from repeated standing-slip training to a novel gait-slip in older ballet dancers compared to non-dancers.

5.4 Conclusions

Overall, this study suggests that older ballet dancers experience fewer falls, have better stability control, take a more effective recovery step, and activate key lower extremity muscles faster than non-dancers of comparable age and sex when exposed to an unexpected slip during treadmill standing. The ballet dancers also adapted faster to the repeated standing-slips and improved their faller rate by the final slip more than the non-dancers. More specifically, the change in stability at recovery step liftoff from first to final standing-slip indicated a more effective reactive adaptation in the dancers compared to the non-dancers, but a similar proactive adaptation between the two groups. Lastly, the ballet dancers showed more transfer of the fall resistant skills learned from the standing-slips to an unexpected and novel gait-slip than their non-dancer counterparts. The dancers exhibited better dynamic gait stability at recovery foot touchdown and shorter lower extremity muscle latencies, which both contributed to their lower gait-slip faller rate. The findings suggest that older adults who practice ballet show better reactions to a novel standing-slip, faster adaptation to repeated standing-slips, and greater transfer of the fall avoidance skills acquired from the standing-slips to a gait slip than those who do not practice ballet. Collectively, the results showed strong associations between the neurobiomechanical mechanisms of ballet practice and reduced fall risk for older adults.

5.5 Implications

This was the first study to expose older ballet dancers to slip perturbations. The findings furnish meaningful evidence that ballet practice is associated with a decrease in the risk of slip-induced backward balance loss and falls in older adults. These findings concur with previous studies reporting that dance-based interventions have successfully improved balance (Blanco-Rambo et al., 2022; Dos Santos Delabary et al., 2018; Harrison et al., 2024; Liu et al., 2021; Sharp and Hewitt, 2014) and functional mobility (Dos Santos Delabary et al., 2020; Haussler and Earhart, 2023) in older populations. Additionally, the current study could enrich our understanding of the underlying mechanisms of ballet practice improving resilience to backward balance loss from biomechanical and neuromuscular perspectives. The more effective recovery stepping, better trunk movement control, and faster leg muscle activation of the older ballet dancers resulted in better stability and a decreased fall risk in response to a slip perturbation compared to the non-dancers (Figure 5.1). It is possible that these favorable observations in the dancers are due specifically to their experience with ballet training. A thoroughly understood mechanism of ballet practice reducing falls will afford a theoretical foundation for developing ballet-based fall prevention programs.

5.6 Limitations and Further Research

This study has limitations that should be addressed in future work in this line of research. First, the participants in this study were all healthy older adults. Thus, it remains unknown how ballet practice or ballet-based interventions may affect dynamic gait stability and fall risk following a slip in older adults with movement disorders (such as Parkinson's disease, multiple sclerosis, stroke, etc.) or cognitive impairments. Given that these neurological conditions heighten the fall risk in older adults (Mahmoudzadeh Khalili et al., 2024; O'Malley et al., 2022; Simpkins et al., 2024; Ullrich et al., 2023), it is meaningful to investigate if and how ballet practice could also lead to fall resistant abilities in these populations. Second, most of the participants in this study were female (85% female dancers and 83% female non-dancers), so the findings may not be fully generalizable to both sexes. However, this imbalanced ratio of females to males is somewhat typical of adult ballet classes according to many ballet schools (e.g., Atlanta Ballet, Motus Dance, and Dance 101), thus the current study's sample of older ballet dancers could reflect this specialized population. Third, this study only examined the EMG latency of four muscles in the lower extremity, so it is still unclear if the EMG signals of other muscles (such as the trunk) differ following standing-slips and gait-slips between older ballet dancers and non-dancers. Furthermore, other EMG metrics such as the burst, peak, and integrated EMG were also not investigated in the current study and should be examined in future work.

Fourth, this study only investigated the responses to standing-slips and gait-slips. Therefore, other types of perturbations such as standing-trips and gait-trips on the treadmill or overground have yet to be examined in older adults who practice ballet. Fifth, the primary outcome measure in this study was dynamic gait stability, which describes the fall risk in the horizontal direction. However, falls are a phenomenon ultimately resulting from limb collapse, which is in the vertical direction (Pai et al., 2006; Yang et al., 2009). Limb support is directly related to lower limb strength and power, both of which have been associated with falls (Ding and Yang, 2016; Han and Yang, 2015; Simpkins and Yang, 2022b). Although the leg muscle strengths were measured under isometric conditions in this study, it could be more fruitful to examine the joint moment during the recovery process from the slip, which requires the measurement of ground reaction forces. Sixth, this study only involved lab-simulated falls and did not examine the association between ballet practice in older adults and real-life falls. Although the real-life falls were not significantly different between groups in the current project, the dancers showed a trend with a lower retrospective annual faller rate (15% vs. 22%, Table 4.1). Such information could provide preliminary results for designing future studies to examine how ballet practice may lower the prospective fall risk for older adults.

Finally, the observational case-control design of the current study only allows for an understanding of the association but not any causal relationship between ballet practice and fall risk reduction in older adults. The slip-perturbation testing was highly standardized in this study. However, the ballet training of the dancer participants was not controlled and could vary greatly within the dancer group in terms of duration, frequency, intensity, location, etc. In addition, dancers and non-dancers also participated in various other physical activities. Future studies should utilize an experimental design such as a randomized controlled trial with a standardized ballet training program as an intervention to better understand the potential causal relationships between ballet practice and these fall-related outcomes. Future work should aim to bridge such knowledge gaps to further understand the mechanisms of ballet practice as related to fall risk. In addition, more work is needed to identify the optimal ballet movements by examining their bio-mechanics (Wells and Yang, 2021b; Wells and Yang, 2022), training dosages, program lengths, etc. that can maximize the training effects on reducing falls while minimizing the risk of injury during ballet training. Another possible research topic is to combine ballet practice with other existing training modalities to possibly result in additive effects on reducing the risk of falls for different populations vulnerable to falls.

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APPENDICES

Appendix A: Health and Dance History Questionnaire

Georgia State University – Biomechanics Laboratory
Health/Dance History Questionnaire

Subject ID: ________________ Age: ___________

Sex: Male Female

Gender: Male Female Transgender Non-Binary

Emergency Contact ______________________________ Phone # _________________

<u>Health Information</u>

1. **Have you ever been diagnosed with any of the following conditions?**

Yes______ No__________ If yes, please put approximate year of onset in space provided.

Neuropathies __________________ Other neurological conditions _____________

Osteoporosis___________________ Other movement disorders _________________

Rheumatoid arthritis___________________ Other arthritic conditions __________________

2. Have you ever been diagnosed with/had any of the following conditions?

Yes______ No__________ If yes, please describe what kind.

Joint replacement ____

Uncorrected visual problems ____

3. Do you currently experience any of these symptoms in your legs or feet? Check all that apply.

Numbness _______ Tingling ___________ Arthritis ___________ Swelling __________

4. How would you describe your overall health?

Excellent ____ Very good ____ Good ____ Fair ____ Poor_____________

<u>Dance History</u>

1. **Do you currently take ballet classes?** Yes No

   (If answer is no, skip to Physical Activity History question)

2. **How many days per week do you currently take ballet class? (Circle response)**

   One Two Three Four Five Six Seven

3. **How many minutes long is the typical ballet class that you take? (Circle response)**

   <60 minutes 60 minutes 90 minutes >90 minutes Other: ________________

4. **For how many months have you maintained this training schedule? _______________**

5. **Do you take other kinds of dance classes currently?** Yes No

   If yes, please list dance types and frequency of training:

   ____

   ____

6. **Are you also currently a ballet teacher?** Yes No

7. **If yes, how many days ______ and hours per day _______ do you typically teach ballet?**

8. **Have you had a lower extremity fracture in the past 6 months?** Yes _____ No ___________

   If yes, please list when this occurred and briefly explain the injury (or injuries):

   ____

   ____

<u>Physical Activity History</u>

1. **Do you currently do any other kind(s) of exercise on a regular basis?** Yes No

   If yes, please list what types and approximately how often you do them:

   ____

   ____

Appendix B: Fall History Questionnaire

Questionnaire for Incidence of Real-life Falls in Past Year

Subject ID: ________________ Date: ______________________

Group: Dancer Non-Dancer

1. Did you experience a fall in the past 12 months? Yes No

If YES:

2. How many falls have you had in the past 12 months? ___________________

For each fall:

3. When did you fall? ____

4. Where did you fall? ____

5. What were you doing at the time of the fall?____________________________________

6. What caused your fall? (Check box below)

☐ Slips ☐ Trips

☐ Activities of daily living/ transfers (getting in/out of bed, sit to stand, turning, reaching)

☐ External hazards (run down by motorist/bicyclist, fall from a ladder/stepping stool)

☐ Other (e.g. fainting)

7.Were you injured from your fall? ____

If yes. Please describe your injury: ____

Appendix C: Rapid Assessment of Physical Activity Questionnaire

Rapid Assessment of Physical Activity

Subject ID: _______________ Date: ___________________

Part	Score	Question	Check answer	
			Yes	No
I Aerobic	1	I rarely or never do any physical activities.		
	2	I do some light or moderate physical activities, but not every week.		
	3	I do some light physical activity every week.		
	4	I do moderate physical activities every week, but less than 30 minutes a day or 5 days a week.		
	5	I do vigorous physical activities every week, but less than 20 minutes a day or 3 days a week.		
	6	I do 30 minutes or more a day of moderate physical activities, 5 or more days a week.		
	7	I do 20 minutes or more a day of vigorous physical activities, 3 or more days a week.		
II Strength & Flexibility	1	I do activities to increase muscle strength, such as lifting weights or calisthenics, once a week or more.		
	2	I do activities to improve flexibility, such as stretching or yoga, once a week or more.		

Notes:
Light activities: your heart beats slightly faster than normal & you can talk and sing (e.g.: walking leisurely, stretching, vacuuming or light yard work, etc.);
Moderate activities: your heart beats faster than normal & you can talk but not sing (e.g.: fast walking, aerobics class, strength training, swimming gently, etc.);
Vigorous activities: your heart rate increases a lot & you can't talk or your talking is broken up by large breaths (e.g.: stair machine, jogging or running, tennis, racquetball, pickleball, or badminton, etc.).

Scoring Instructions:
Part I: Aerobic

To score, choose the question with the highest score with an affirmative response.
1 → Sedentary 2→ Under-active
3→ Under-active regular-light activities
4 & 5→ Under-active regular 6 & 7 → Active

Part II: Strength and Flexibility
If "Yes" checked for both, score 3; If "No" checked for both, score 0.

To obtain the total score, add the sub-scores from both parts I and II.

Appendix D: Power Analysis

The sample size for this study was calculated based on results using the primary outcome measure of dynamic gait stability from a pilot study using young professional ballet dancers (Simpkins et al., 2022a). The estimated effect size (*d)* of dynamic gait stability at TD from the independent *t*-test in the pilot study was 0.794. With a two-tailed alpha level of 0.05 and a statistical power of 0.80, the G*Power software (Faul et al., 2007) revealed that 25 subjects per group would be required.

Appendix E: Supplementary Tables

Table A1 The test statistic, effect size, 95% confidence intervals (CI), and *p*-values for all primary and secondary outcome measures for the first standing-slip trial. *Mann-Whitney *U* test was used, and the 95% CI indicates the median difference.

Outcome	Test Statistic	Effect Size	95% CI	*p*-value
Stability				
ON	$Z = 1.632$	$d = 0.514$	[-0.047, 0.004]*	0.052
LO	$F = 6.899$	$d = 0.830$	[-0.353, -0.316]	0.006
TD	$t = -2.361$	$d = 0.722$	[-0.244, -0.019]	0.012
COM position				
ON	$Z = 1.777$	$d = 0.563$	[-0.092, 0.005]*	0.038
LO	$t = -2.191$	$d = 0.670$	[-0.294, -0.012]	0.017
TD	$t = -2.364$	$d = 0.723$	[-0.570, -0.045]	0.012
COM velocity				
ON	$t = 1.366$	$d = 0.417$	[-0.006, 0.003]	0.090
LO	$t = 3.089$	$d = 0.944$	[0.012, 0.060]	0.002
TD	$Z = 0.183$	$d = 0.056$	[-0.030, 0.023]*	0.428
Step length	Z = 0.637	$d = 0.195$	[-0.022, 0.020]*	0.262
Step latency	$t = -2.215$	$d = 0.650$	[-0.527, -0.005]	0.020
Step duration	$Z = -2.294$	$d = 0.747$	[0.000, 0.030]*	0.011
Step speed	$Z = 1.860$	$d = 0.590$	[-0.238, 0.035]*	0.032
Slip distance	$t = 2.270$	$d = 0.694$	[0.003, 0.056]	0.015
Trunk angle				
ON	$F = 1.202$	$d = 0.138$	[-4.427, -2.052]	0.140
LO	$Z = 1.071$	$d = 0.330$	[-5.338, 1.315]*	0.142
TD	$F = 0.165$	$d = 0.127$	[-7.662, -1.958]	0.344
Trunk angular velocity				
ON	$Z = 1.084$	$d = 0.140$	[-0.787, 0.250]*	0.140
LO	$t = 1.019$	$d = 0.312$	[-14.591, 44.335]	0.157
TD	$t = 1.031$	$d = 0.315$	[-15.791, 48.707]	0.155

Table A2 The mean value and standard deviation for all primary and secondary outcome measures from the first standing-slip trial for dancers and non-dancers. A negative trunk angle indicates a backward-leaning trunk. A negative trunk angular velocity indicates a backward-moving trunk. The *p*-values are also provided.

Outcome	**Group**		***p*-value**
	Dancers (*n* = 20)	Non-Dancers (*n* = 23)	
Stability			
ON	0.190 ± 0.042	0.168 ± 0.027	0.052
LO	-0.520 ± 0.095	-0.550 ± 0.063	0.006
TD	-0.217 ± 0.187	-0.348 ± 0.178	0.012
COM position			
ON	0.392 ± 0.080	0.349 ± 0.051	0.038
LO	-0.601 ± 0.244	-0.753 ± 0.241	0.017
TD	-0.310 ± 0.396	-0.617 ± 0.449	0.012
COM velocity			
ON	-0.009 ± 0.002	-0.008 ± 0.003	0.090
LO	-0.243 ± 0.031	-0.207 ± 0.043	0.002
TD	-0.067 ± 0.030	-0.070 ± 0.048	0.428
Step length (/*bh*)	0.125 ± 0.039	0.119 ± 0.034	0.262
Step latency (s)	0.344 ± 0.053	0.378 ± 0.062	0.020
Step duration (s)	0.130 ± 0.024	0.142 ± 0.016	0.011
Step speed (*bh*/s)	0.995 ± 0.357	0.838 ± 0.238	0.032
Slip distance (m)	0.258 ± 0.045	0.288 ± 0.040	0.015
Trunk angle (deg)			
ON	-2.620 ± 3.534	-3.896 ± 3.875	0.140
LO	-2.425 ± 5.720	-5.234 ± 5.887	0.142
TD	-4.104 ± 10.325	-5.269 ± 8.744	0.344
Trunk angular velocity (deg/s)			
ON	0.429 ± 0.875	0.134 ± 0.605	0.140
LO	-12.882 ± 51.658	1.990 ± 44.031	0.157
TD	10.041 ± 54.790	26.499 ± 49.912	0.155

Table A3 The test statistic, effect size, 95% confidence intervals (CI), and *p*-values for all electromyography latencies for the first standing-slip trial. Values for both the recovery stepping leg and the standing leg are provided. *Mann-Whitney *U* test was used, and the 95% CI indicates the median difference.

Muscle Latency	**Test Statistic**	**Effect Size**	**95% CI**	***p*-value**
Rectus femoris				
Stepping	$t = 0.737$	$d = 0.247$	[-0.009, 0.019]	0.233
Standing	$t = 1.980$	$d = 0.661$	[-0.005, 0.035]	0.056
Biceps femoris				
Stepping	$t = 1.934$	$d = 0.628$	[-0.001, 0.032]	0.061
Standing	$t = 1.204$	$d = 0.377$	[-0.008, 0.030]	0.118
Tibialis anterior				
Stepping	$t = 2.203$	$d = 0.719$	[0.001, 0.029]	0.017
Standing	$t = 3.156$	$d = 1.003$	[0.009, 0.039]	0.002
Medial gastrocnemius				
Stepping	$t = 1.969$	$d = 0.732$	[-0.002, 0.075]	0.030
Standing	$Z = 0.206$	$d = 0.076$	[-0.068, 0.052]*	0.420

Table A4 The mean value and standard deviation for all electromyography latencies (in seconds) for the first standing-slip trial. Values for both the recovery stepping leg and the standing leg are provided. The *p*-values and sample sizes are also provided.

Muscle Latency	**Group**				***p*-value**
	Dancers		Non-Dancers		
	Mean ± SD	*n*	Mean ± SD	*n*	
Rectus femoris					
Stepping	0.146 ± 0.017	16	0.151 ± 0.023	20	0.233
Standing	0.141 ± 0.023	17	0.158 ± 0.029	19	0.028
Biceps femoris					
Stepping	0.121 ± 0.028	18	0.137 ± 0.021	20	0.031
Standing	0.124 ± 0.023	19	0.136 ± 0.034	22	0.118
Tibialis anterior					
Stepping	0.116 ± 0.017	17	0.131 ± 0.024	21	0.017
Standing	0.119 ± 0.021	18	0.143 ± 0.026	22	0.002
Medial gastrocnemius					
Stepping	0.190 ± 0.045	14	0.227 ± 0.055	15	0.030
Standing	0.206 ± 0.113	10	0.212 ± 0.081	19	0.434

Table A5 The test statistic, effect size, 95% confidence intervals (CI), and *p*-values for all primary and secondary outcome measures for the percent change from the first standing-slip to the last standing-slip. *Mann-Whitney *U* test was used, and the 95% CI indicates the median difference.

Outcome	Test Statistic	Effect Size	95% CI	*p*-value
Stability				
ON	t = 0.636	d = 0.194	[-0.098, 0.189]	0.264
LO	F = 6.436	d = 0.720	[0.176, 0.259]	0.007
TD	Z = 1.485	d = 0.464	[-1.120, 0.170]*	0.069
COM position				
ON	t = 0.741	d = 0.226	[-0.082, 0.178]	0.232
LO	F = 0.786	d = 0.278	[2.012, 2.372]	0.191
TD	Z = 0.450	d = 0.138	[-1.250, 0.620]*	0.326
COM velocity				
ON	Z = 0.731	d = 0.223	[-0.470, 0.150]*	0.233
LO	F = 10.939	d = 0.972	[-0.275, -0.119]	0.001
TD	Z = 1.815	d =0.577	[-0.810, 0.010]*	0.035
Step length	Z = -0.037	d = 0.012	[-0.210, 0.310]*	0.486
Step latency	F = 0.496	d = 0.220	[-0.318, -0.258]	0.243
Step duration	t = -2.815	d = 0.861	[-0.393, -0.065]	0.004
Step speed	Z = -2.315	d = 0.755	[0.063, 0.500]*	0.011
Slip distance	Z = -2.352	d = 0.769	[0.020, 0.160]*	0.010
Trunk angle				
ON	Z = -0.487	d = 0.148	[-0.380, 0.730]*	0.313
LO	Z = 0.730	d = 0.223	[-1.020, 0.590]*	0.233
TD	Z = 1.522	d = 0.335	[-1.570, 0.190]*	0.064
Trunk angular velocity				
ON	Z = 0.438	d = 0.134	[-0.990, 0.610]*	0.331
LO	Z = 2.301	d = 0.750	[-2.940, -0.230]*	0.011
TD	Z = 2.898	d = 0.985	[-2.270, -0.380]*	0.002

Table A6 The mean value and standard deviation for the percent change from first standing-slip to last standing-slip of all primary and secondary outcome measures for dancers and non-dancers. The *p*-values are also provided.

Outcome *All presented as percent change*	**Group**		***p*-value**
	Dancers (*n* = 20)	Non-Dancers (*n* = 23)	
Stability			
ON	13.285 ± 16.612	17.781 ± 27.880	0.264
LO	49.020 ± 21.889	34.822 ± 17.276	0.007
TD	249.423 ± 351.613	140.688 ± 228.213	0.069
COM position			
ON	12.193 ± 14.783	16.823 ± 25.416	0.232
LO	32.729 ± 253.475	67.373 ± 28.361	0.191
TD	295.318 ± 544.605	254.885 ± 628.045	0.326
COM velocity			
ON	-5.810 ± 35.458	-26.649 ± 67.790	0.233
LO	-4.350 ± 21.858	-34.752 ± 27.980	0.001
TD	72.078 ± 46.096	-36.912 ± 248.010	0.035
Step length	25.080 ± 52.594	32.711 ± 55.205	0.486
Step latency	-29.945 ± 11.755	-28.264 ± 9.881	0.243
Step duration	24.363 ± 29.370	1.510 ± 23.849	0.004
Step speed	2.270 ± 42.116	30.119 ± 38.337	0.011
Slip distance	-44.776 ± 15.269	-37.595 ± 14.198	0.010
Trunk angle			
ON	17.609 ± 198.103	47.124 ± 135.820	0.313
LO	49.113 ± 134.012	25.731 ± 144.443	0.233
TD	-124.681 ± 263.451	-105.448 ± 127.890	0.064
Trunk angular velocity			
ON	31.072 ± 129.816	17.863 ± 156.285	0.331
LO	127.421 ± 254.762	-92.477 ± 260.107	0.011
TD	-41.095 ± 203.848	-178.906 ± 167.996	0.002

Table A7 The number of fallers and percentage of fallers for all 16 standing-slip trials in the dancer group and the non-dancer group. The *p*-values are also provided.

Standing-slip Trial	**Group**		***p*-value**
	Dancers	**Non-Dancers**	
1	9 (45%)	19 (83%)	0.005
2	9 (45%)	16 (70%)	0.052
3	6 (30%)	14 (61%)	0.022
4	4 (20%)	14 (61%)	0.004
5	3 (15%)	13 (57%)	0.003
6	5 (25%)	15 (65%)	0.004
7	4 (20%)	14 (61%)	0.004
8	4 (20%)	15 (65%)	0.002
9	5 (25%)	15 (65%)	0.004
10	4 (20%)	13 (57%)	0.008
11	4 (25%)	14 (64%)	0.010
12	3 (19%)	14 (64%)	0.003
13	2 (13%)	12 (55%)	0.004
14	2 (13%)	13 (59%)	0.003
15	1 (7%)	13 (59%)	< 0.001
16	1 (7%)	13 (59%)	< 0.001

Table A8 The test statistic, effect size, 95% confidence intervals (CI), and *p*-values for the percent change from first to last standing-slip for all electromyography latencies. Values for both the recovery stepping leg and the standing leg are provided. *Mann-Whitney *U* test was used, and the 95% CI indicates the median difference.

Muscle Latency *All presented as percent change*	**Test Statistic**	**Effect Size**	**95% CI**	***p*-value**
Rectus femoris				
Stepping	t = 2.775	d = 0.848	[0.028, 0.180]	0.004
Standing	t =1.099	d = 0.336	[-0.027, 0.090]	0.139
Biceps femoris				
Stepping	t = -1.287	d = 0.394	[-0.219, 0.048]	0.103
Standing	t = 0.689	d = 0.211	[-0.086, 0.176]	0.247
Tibialis anterior				
Stepping	Z = -1.694	d = 0.534	[-0.010, 0.200]*	0.045
Standing	t = -0.093	d = 0.028	[-0.112, 0.102]	0.463
Medial gastrocnemius				
Stepping	t = -0.974	d = 0.298	[-0.192, 0.067]	0.168
Standing	Z = -0.707	d = 0.198	[-0.220, 0.430]*	0.260

Table A9 The mean value and standard deviation for the percent change from the first standing-slip to the last standing-slip of all electromyography latencies. Values for both the recovery stepping leg and the standing leg are provided. The *p*-values are also provided.

Muscle Latency *All presented as percent change*	**Group**		***p*-value**
	Dancers (n = 20)	Non-Dancers (n = 23)	
Rectus femoris			
Stepping	-4.930 ± 9.656	5.459 ± 14.134	0.004
Standing	-5.512 ± 20.037	0.430 ± 18.445	0.139
Biceps femoris			
Stepping	-0.288 ± 24.745	-8.852 ± 18.765	0.103
Standing	-6.250 ± 16.050	-1.768 ± 24.935	0.247
Tibialis anterior			
Stepping	-8.211 ± 14.653	6.548 ± 27.587	0.045
Standing	-2.493 ± 15.362	-3.041 ± 18.928	0.463
Medial gastrocnemius			
Stepping	-5.756 ± 22.045	-11.919 ± 20.065	0.168
Standing	33.268 ± 54.616	44.767 ± 59.327	0.260

Table A10 The test statistic, effect size, 95% confidence intervals (CI), and *p*-values for all primary and secondary outcome measures for the gait-slip trial. *Mann-Whitney *U* test was used, and the 95% CI indicates the median difference.

Outcome	Test Statistic	Effect Size	95% CI	*p*-value
Stability				
LTD	Z = -0.048	d = 0.016	[-0.034, 0.043]*	0.488
RLO	F = 1.295	d = 0.397	[0.058, 0.095]	0.132
RTD	Z = 2.262	d = 0.814	[-0.393, -0.035]*	0.012
COM position				
LTD	t = -0.072	d = 0.024	[-0.112, 0.105]	0.437
RLO	F = 1.328	d = 0.403	[-0.198, -0.124]	0.129
RTD	Z = 2.647	d = 0.983	[-0.671, 0.115]*	0.004
COM velocity				
LTD	t = -1.226	d = 0.415	[-0.010, 0.002]	0.114
RLO	Z = -0.032	d = 0.010	[-0.008, 0.007]*	0.488
RTD	Z = 0.786	d = 0.264	[-0.097, 0.034]*	0.223
Step length	t = -3.161	d = 1.069	[-0.402, -0.087]	0.002
Step duration	t = 0.785	d = 0.265	[-0.038, 0.085]	0.219
Step speed	t = -2.496	d = 0.844	[-0.205, -0.021]	0.009
Slip distance	F = 1.428	d = 0.414	[-0.736, -0.685]	0.121
Trunk angle				
LTD	Z = 0.530	d = 0.177	[-2.922, 1.833]*	0.306
RLO	Z = 0.915	d = 0.310	[-3.151, 1.490]*	0.187
RTD	t = -0.783	d = 0.265	[-4.505, 2.000]	0.220
Trunk angular velocity				
LTD	F = 0.391	d = 0.220	[13.550, 18.520]	0.268
RLO	Z = 0.594	d = 0.199	[-5.615, 3.323]*	0.284
RTD	t = 2.412	d = 0.815	[5.565, 65.206]	0.011

Table A11 The mean value and standard deviation for all primary and secondary outcome measures from the gait-slip trial for dancers and non-dancers. A negative trunk angle indicates a forward-leaning trunk. A negative trunk angular velocity indicates a forward-moving trunk. The *p*-values are also provided.

Outcome	**Group**		***p*-value**
	Dancers (*n* = 15)	Non-Dancers (*n* = 21)	
Stability			
LTD	-0.151 ± 0.037	-0.153 ± 0.063	0.488
RLO	0.084 ± 0.052	0.071 ± 0.055	0.132
RTD	0.290 ± 0.265	0.081 ± 0.166	0.012
COM position			
LTD	-0.659 ± 0.107	-0.663 ± 0.186	0.437
RLO	-0.147 ± 0.108	-0.169 ± 0.119	0.129
RTD	0.525 ± 0.424	0.151 ± 0.274	0.004
COM velocity			
LTD	0.192 ± 0.007	0.188 ± 0.010	0.114
RLO	0.209 ± 0.010	0.207 ± 0.016	0.488
RTD	0.036 ± 0.100	0.010 ± 0.081	0.223
Step length (/*bh*)	0.134 ± 0.074	0.051 ± 0.083	0.002
Step duration (s)	0.451 ± 0.080	0.478 ± 0.098	0.219
Step speed (*bh*/s)	0.288 ± 0.136	0.173 ± 0.132	0.009
Slip distance (m)	0.185 ± 0.033	0.209 ± 0.043	0.121
Trunk angle (deg)			
LTD	-0.283 ± 3.577	-0.688 ± 4.057	0.306
RLO	1.257 ± 3.615	0.726 ± 4.067	0.187
RTD	3.736 ± 4.992	2.483 ± 4.545	0.220
Trunk angular (deg/s)			
LTD	16.444 ± 8.240	15.972 ± 7.522	0.268
RLO	-1.150 ± 5.558	-1.086 ± 6.856	0.284
RTD	5.115 ± 43.614	40.500 ± 43.258	0.011

Table A12 The test statistic, effect size, 95% confidence intervals (CI), and *p*-values for all electromyography latencies of the recovery stepping leg for the gait-slip trial. *Mann-Whitney *U* test was used, and the 95% CI indicates the median difference.

Outcome	**Test Statistic**	**Effect Size**	**95% CI**	***p*-value**
Rectus Femoris	$t = 1.930$	$d = 0.717$	[-0.004, 0.141]	0.032
Biceps Femoris	$Z = -1.886$	$d = 0.661$	[-0.003, 0.080]*	0.031
Tibialis Anterior	$t = 1.775$	$d = 0.724$	[-0.014, 0.185]	0.045
Medial Gastrocnemius	$t = 1.846$	$d = 0.686$	[-0.006, 0.123]	0.038

Table A13 The mean value and standard deviation for all electromyography latencies (in seconds) for the stepping leg of the gait-slip trial. The *p*-values and sample sizes are also provided.

Muscle Latency	**Group**				***p*-value**
	Dancers		Non-Dancers		
	Mean ± SD	*n*	Mean ± SD	*n*	
Rectus femoris	0.401 ± 0.100	14	0.469 ± 0.090	15	0.032
Biceps femoris	0.330 ± 0.040	13	0.375 ± 0.071	16	0.031
Tibialis anterior	0.367 ± 0.120	12	0.452 ± 0.116	12	0.045
Medial gastrocnemius	0.407 ± 0.073	14	0.466 ± 0.095	15	0.038