Detecting differences in gait initiation between older adult fallers and non-fallers through time-series principal component analysis (PCA)

by

Kaya Yoshida
BSc (Honours), University of Victoria, 2020

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We acknowledge and respect the lək̓ʷəŋən peoples on whose traditional territory the university stands and the Songhees, Esquimalt and W̱SÁNEĆ peoples whose historical relationships with the land continue to this day.

Supervisory Committee

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Supervisory Committee

Dr. Sandra Hundza, School of Exercise Science Physical and Health and Education

Co-Supervisor

Dr. Marc Klimstra, School of Exercise Science Physical and Health and Education

Co-Supervisor

Abstract

Gait initiation (GI) is an important locomotor transition task that includes anticipatory postural adjustments and the joint propulsion necessary for the first step of walking. Metrics associated with this task are known to change across the lifespan and may provide valuable information for fall risk indication, as falls often occur during transitional tasks. Assessments of discrete variables between fallers and non-fallers at GI have provided insight into differences between groups. However, more complex approaches such as time-series principal component analysis (PCA) may allow the examination of changes in magnitude, pattern, and timing not detectable using discrete comparisons alone. Therefore, this thesis aims to characterize differences between fallers and non-fallers by examining the kinematics and kinetics of gait initiation using time-series PCA. A sample of 56 community-dwelling older adults was recruited for this study and completed five walking trials where GI was measured by two force platforms. PCA of centre of pressure kinematics and kinetics time-series data were used to identify the critical features of the signal, and multivariate analysis of covariance was used to compare the individual loading scores of each principal component for each phase between groups. It was revealed that fallers demonstrated differences in the range of mediolateral movement during weight transfer and forward progression, a greater range of anteroposterior movement in forward progression, and a more gradual rise in vertical forces in the first step, associated with a shorter first step length. These findings point to a tendency for fallers to prioritize stability over forward progression performance, and differences in postural control strategies, compared to non-fallers. Further, the use of time-series PCA helped to highlight differences not detectable using discrete analysis alone.

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Dedication

To my family, whose unconditional support has allowed me to be where I am today. My dad, for instilling in me his love of science, and my mom for always believing in me. My sister, who reminds me to love what I do, and keeps a smile on my face every day. My wonderful friends, who lift me up, and constantly cheer me on. I appreciate each and every one of you!

Chapter 1 — Background Information and Literature Review

Operational Definitions

Antero-posterior (AP): Referring to the direction going from front to back, or vice versa.

Anticipatory postural adjustments (APA): muscle activation and postural sway before a forthcoming perturbation, such as the onset of movement

Base of support (BoS): The area within the lines connecting the outer perimeter of each of the points of support

Center of Mass (COM): The average location of mass concentration within a rigid body (Robertson et al., 2004).

Center of Pressure (COP): The location of the force vector resulting from the distribution of forces on a contact area, reflective of weight shift and muscular control (Chang & Krebs, 1999; Robertson et al., 2004)

Community-dwelling older adults: Adults aged 65 years or older, residing independently and with the absence of any chronic health conditions.

Fall: “An event which resulted in a person coming to rest unintentionally on the ground or other lower level, not as the result of a major intrinsic event or an overwhelming hazard” (Lord et al., 1994).

Gait initiation (GI): The transitional phase between static balance in an upright position and the start of walking (Chang & Krebs, 1999). GI also encompasses anticipatory postural adjustments (APA) and the propulsive joint forces necessary to generate the first step of walking gait (Winter, 2009).

Gait variability: The fluctuations of gait characteristics from one step to the next

Gait: The unique characteristics of a person's walking

Ground reaction force (GRF): The reaction force exerted by the ground to the body in contact with it

Individual loading score (ILS): The score for each subject on a given principal component

Medio-lateral (ML): Referring to the direction going from side to side, in the frontal plane

Principal component analysis (PCA): A orthogonal waveform decomposition technique resulting in independent PCs to describe the maximum amount of variance within the original dataset. A dimensional reduction technique to interpret and analyze temporal waveform data (Robertson et al., 2004).

Principal component (PC): The linear combinations explaining the most variance of the initial variables after computing a PCA

Step: The heel-to-heel or toe-to-toe distance from the ipsilateral to the contralateral side of consecutive steps during walking gait

Stride: The heel-to-heel or toe-to-toe distance of the same foot during consecutive steps during walking gait

Assumptions and Limitations

In this study, it is assumed that: participants will accurately and honestly disclose all relevant information to the specified inclusion/exclusion criteria; participants accurately report their fall history, including relevant details to classify their fall; each participant completes the walking gait protocol to the best of their physical ability.

This study is limited by its sample of community-dwelling older adults currently residing in the Greater Victoria region. This study assumes that these results will have some generalizability to the greater population of community-dwelling older adults in North America but recognizes that selection bias of relatively healthy older adults limits the application of the results in the broader population of older adults.

Falls Among Older Adults

Prevalence of Falls in Canada

Canada has seen a dramatic shift in the age of the average citizen; in recent years, the number of adults aged 65 and older surpassed the number of children; a gap that continues to widen, with older adults now comprising 18% of the population (Statistics Canada, 2020). Of those older adults, the Public Health Agency of Canada (2020) estimates that between 20-30% of seniors will experience a fall annually. This increasing demographic points to a resultant increase in individuals who present as a risk for falling, and thus a need for means to detect and mitigate risk factors associated with falling.

Consequences of Falls

Falls are associated with higher incidences of morbidity, mortality and inflict substantial costs to health care. Specifically, falls and injuries related to falls are the leading cause of mortality after injury in older adults (Rubenstein, 2006). Falls are a significant threat to the independence of older adults and contribute to the marked decrease in quality of life among the aging population. Falls often precede admittance to long-term care homes and result in decreased mobility and independence in older adults due to injury and fear of falling again (Lusardi et al., 2017; Murphy et al., 2003; Scott et al., 2010). Further, the economic costs associated with falls in Canada are estimated to be greater than $7 billion annually in direct and indirect health care costs (SMARTRISK, 2009). Therefore, there is an increased need for sensitive and specific fall risk assessment paired with interventions targeted to vulnerable populations, to prevent these social and economic costs from increasing.

Risk Factors for Falls

Falls are a multi-faceted issue resulting from a combination of intrinsic, extrinsic and environmental factors such as deficits in muscular strength (Daly et al., 2013; Kabeshova et al., 2014), gait (Verghese et al., 2009), balance (Cuevas-Trisan, 2019), sensory input (Dhital et al., 2010), cardiorespiratory function (Mertz et al., 2010) and executive function of cognition (Lusardi et al., 2017; Parihar et al., 2013; Rubenstein, 2006). A systematic review by Deandrea et al. (2010) noted that out of sociodemographic, mobility, sensory, psychological, and medical factors and medication use, the strongest correlate of fall risk among adults aged 65 years and older was a history of falls. This finding is well accepted in the literature; however, it cannot predict first-time fall risk. Following this risk factor, gait issues are the next leading predictor, a finding well corroborated in the literature, along with issues related to dynamic balance and postural control (Ambrose et al., 2013; Deandrea et al., 2010; Lusardi et al., 2017).

Assessments of walking gait helps us understand the expected changes with healthy aging and how as they become impaired may lead to falls (Bayot et al., 2018).

Transitional movements, such as moving from standing to walking, known as gait initiation may also be associated with higher fall risk among older adults. The change from a stable base of support to instability in locomotion may highlight aspects of balance dysfunction and deficits in neuromuscular control, as postural control is challenged. This challenge is more evident among older adults, especially as they become frailer (Chang & Krebs, 1999; Hass et al., 2005, 2008). Falls are a multi-faceted issue and novel methods for understanding the biomechanical basis of fall risk are necessary, and gait initiation assessment may provide novel insights into this issue.

Gait Initiation

Gait initiation (GI) is an important locomotor transition task that is characterized by anticipatory postural adjustments (APA) and the propulsive joint forces necessary to generate the first step of walking gait (Winter, 2009). Understanding APAs and GI allow insight into postural control, stability and dynamic balance important to locomotion in everyday life; they may act as an indicator of neuromuscular control and overall health, and potentially fall risk among older adults (Bayot et al., 2018; Khanmohammadi et al., 2015).

In typical GI, first, the center of pressure (COP) is displaced posteriorly and laterally towards the initial swing limb. This initial COP shift about the initial swing limb allows the generation of forces to propel the body toward the stance limb and forward for the first step. The body's center of mass (COM) is accelerated anteriorly and toward the initial stance limb in reaction to this force (Halliday et al., 1998; Winter, 2009). Literature has described this net COP displacement as an uncoupling of the body from the COM generating the necessary forward momentum and instability necessary for locomotion (Lee et al., 2019; Winter, 2009). Next, as the initial swing limb is picked up off the ground, the body's base of support is now limited to the stance limb, and this instability allows the forward momentum to drive the center of mass forward. The initial swing limb contacts the ground and gait initiation is terminated, as steady-state walking begins (Winter, 2009). The typical center of pressure path during GI is illustrated in Figure 1.1 below.

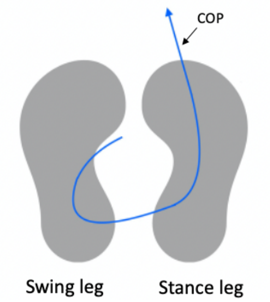


Figure 1.1. The general pattern of center of pressure path during gait initiation when stepping with the left foot.

Gait initiation can be broken down into characteristic phases to better understand the biomechanical demands. These four main components include (1) premovement, (2) weight transfer, (3) forward progression, and (4) the first step.

(1) The premovement stage marks the onset of the APA phase and is often characterized by the oscillations in the COP position, leading up to the beginning of the lateral weight transfer toward the initial stance leg.

(2) During the weight transfer stage, there is a visible shift in the body's COP posteriorly and laterally toward the swing limb, then the stance limb to allow for mobility of the initial stepping leg (Yiou et al., 2017). The initial swing leg is picked up and unilateral stance is achieved, challenging stability in the mediolateral direction, due to the decreased width of the base of support.

(3) Forward progression shows the anterior progression of the COM and COP under the stance limb. As the COM moves toward the boundary of the base of support, mobility is increased to shift the weight of the body forward to be caught by the landing of the initial swing leg. The weight is fully accepted by the stance leg, as the swing phase is carried out. The swing phase includes the time between toe-off of the initial swing leg, until heel strike (Bonora et al., 2015).

(4) Lastly, the first step is characterized initially by the heel strike of the initial swing leg.

Anticipatory postural adjustments as part of GI have been characterized into functional roles based on the axis in which they are observed. For example, along the anteroposterior axis, APAs tend to be predictive of motor performance (Brenière et al., 1987). The COP shifting backwards allows the generation of forces anteriorly to propel the body forward and thus is related to forward progression velocity and step length (Brenière & Do, 1991; Lepers & Brenière, 1995). Along the mediolateral axis, APAs appear to be more related to postural stability (McIlroy & Maki, 1999; Yiou et al., 2012), aiming to reduce the instability created from the separation of the COP and COM at foot-off on the swing side. Increased instability during this time leads to an increased risk of falls (Yiou et al., 2012). Falls in the mediolateral direction are more common in older adults, compared to falls in the anteroposterior direction. They are also associated with a six-fold greater risk of hip fracture in that direction, presenting a significant healthcare concern (Maki et al., 2000; Parkkari et al., 1999). Thus, changes in postural control may be observable in APAs at the onset of walking gait, making this task important to consider when assessing fall risk among older adults.

Changes in Gait Initiation Across the Lifespan

During normal, healthy aging, several well-understood changes occur across a wide variety of dimensions of human health. This multidimensional depreciation of functional ability has far-reaching implications in terms of quality of life, the performance of daily activities, and injury risk, and is evident in gait initiation (Halliday et al., 1998; Muir et al., 2014). Across the lifespan, the motor program from which GI is expressed remains fairly robust (Polcyn et al., 1998). However, there are noticeable changes that occur between young adults and healthy older adults. During GI, the displacement of COP posteriorly decreases as age increases, limiting one’s ability to accelerate the COM forward (Chang & Krebs, 1999). Thus, the aging neuromuscular system faces challenges during GI, requiring a higher division of attentional resources to mitigate these challenges. Indeed, GI has been found to require higher attentional resources, measured by increases in cortical activation within the prefrontal cortex, premotor area and supplementary motor areas during the onset of locomotion, compared to steady-state walking (Suzuki et al., 2004).

In assessments of GI in healthy populations of both younger and older adults, differences have been revealed in terms of reduced amplitudes of muscle activation in the leading limb among the older cohort. This is paired with a lesser and slower backward displacement of COP and may point to an increased risk of fall, specifically in transitional tasks such as GI (Khanmohammadi et al., 2015). The apparent decrease in force generation with increasing age has been widely corroborated across gait initiation literature (Muir et al., 2014; Patchay et al., 2002; Satoh et al., 2019). Healthy older adults have significantly slower gait speed during the first step of initiation by up to 35% when compared to healthy young adults; however, this was specific to the first step, and not consistently observed in steady-state walking. This has been attributed to deficits in momentum generation, specific to GI among healthy older adults (Muir et al., 2014). Moreover, across all age groups, the widest step was consistently the first, indicative of a protective strategy to account for the transitional instability due to ML COP displacement following weight transfer into unilateral stance (Muir et al., 2014).

Gait Initiation as an Indicator of Health

A variety of pathologies, specifically affecting the central nervous system and musculoskeletal systems are known to expedite the effects of the normal aging process or create other adverse effects on human function and health. It has been suggested that GI may be useful in characterizing pathological conditions compared to healthy samples. This includes a variety of conditions including Parkinson’s Disease, Marche à petit pas, and acutely post-concussion, (Bonora et al., 2015, 2017; Hass et al., 2008; Mancini et al., 2016; Muniz et al., 2012). As GI is comprised of very characteristic patterns observed in both the COP and COM, it has been used to observe changes in clinical populations compared to healthy controls.

Parkinson's disease

Due to the pathological presentation of Parkinson's Disease (PD), including freezing gait and deficits in postural control, it has been well documented in the literature as it pertains to GI. It has been found that PD patients have significantly smaller COM-COP distances when compared to both healthy younger adults, and healthy older adults at GI. This has been attributed to the more conservative step initiation strategies and a decrease in force production among the aging and pathological samples. As the COM-COP distance increases, the demand for dynamic postural control increases. As such, those with a lower ability to compensate for these perturbations likely adopt more conservative approaches (Martin et al., 2002). Moreover, the lesser generation of forward momentum during the beginning of walking may be related to the shorter steps commonly observed among more frail older adults (Hass et al., 2005; Mbourou et al., 2003; Muir et al., 2014). More recently, using measurements derived from COM trajectories, it was found that differences in APA metrics can be used to discriminate between PD patients and healthy controls, both temporally and spatially. Specifically, a reduction in amplitude of ML trunk acceleration across all phases of GI was observed in the PD group, as well as a longer duration of the weight transfer phase (Bonora et al., 2017).

Postural control and acute-post concussion

It has been found that subjects' post-concussive GI assessments differed from baseline measurements, indicating that postural control may be impaired acutely. These deficits were apparent through a decreased posterior and lateral COP displacement, resulting in ineffective force generation to propel the body forward into walking gait (Buckley et al., 2017; Doherty et al., 2017). Notably, this reduction in COP displacement in the lateral and posterior direction is similar to that observed in Parkinson's Disease, as described above. It is suggested that impairment acutely post-concussion is related to motor planning and execution in the premotor and supplementary motor areas (Buckley et al., 2017). This demonstrates the interconnected nature of postural control between supraspinal elements and biomechanical output and adds to the growing body of literature suggesting that GI may provide indications of changes in postural control and damage to supraspinal elements.

Gait Initiation and Fall Risk

Notably, changes in GI have been linked to fall risk in older adults. Falling older adults are found to have significantly shorter first steps, and up to twice as much variability in first step length compared to non-falling counterparts. These findings were specific to falling older adults and not observed among non-falling older adults (Mbourou et al., 2003). Similarly, in tests of choice stepping and reaction time, a task requiring individuals to rapidly trigger and execute a step, older adults at risk of falls had an increase in APA latency, and more APA errors compared to non-falling counterparts (Lord & Fitzpatrick, 2001; Patla et al., 1993; St George et al., 2007). Further, when APAs are produced inappropriately, there was a clear link to increased risk of falls (Ellmers et al., 2020).

According to Callisaya et al., (2016), under both dual- and single-task conditions, gait initiation time was slower among adults at risk of multiple falls. Specifically, the time to weight transfer was the most strongly associated with fall risk. Further, in the weight transfer phase specifically, an increase in reaction time by 50% has been observed in older adults, compared to healthy young adults, as well as a concurrent decrease in peak vertical force, indicative of less effective motor patterns in the elderly (Patla et al., 1993). This delay in temporal components of step initiation on such tasks have previously been identified as a predictor of fall risk (St George et al., 2007), however, this has currently been limited to choice stepping tasks, and it is unclear to what extent this can be extrapolated to normal, daily life gait initiation.

From these findings, it is clear that aging is associated with changes to postural control, dynamic balance and APAs. Measurements of GI are sensitive enough to detect these changes associated with healthy aging, as well as those potentially associated with fall risk. Further, it has been demonstrated that these changes in postural control are detectable before falls occur, making GI assessments an important consideration to understand the risk of falls with increasing age (Michalska et al., 2021). However, the research into all the specific kinetics and kinematics related to gait initiation and these changes in postural to characterize fallers from non-fallers by phase of GI is lacking. Thus, a clear profile of gait initiation differences between fallers and non-fallers is necessary to wholly understand the differences common among fallers, compared to those who do not experience falls.

Assessment of Fall Risk

Clinical Balance and Mobility Tests

Currently, researchers and clinicians use several approaches to assess fall risk in the elderly. These measures include gait and clinical balance assessments, as well as qualitative methods to understand other risk factors. To date, the most common clinically implemented tests to screen for risk of falls include primarily measurements of mobility and balance. These tests commonly involve measures of balance, muscular strength and endurance, as well as some characteristics of gait (Balasubramanian, 2015; Lusardi et al., 2017; Murphy et al., 2003). The American Geriatric Society (AGS) and British Geriatric Society (BGS) recommend including an evaluation of gait and balance, and suggest tests such as the Timed Up and Go test (TUGT) (Podsiadlo & Richardson, 1991), the Berg Balance Scale (BBS) (Berg et al., 1989), and the Performance-Oriented Mobility Assessment (POMA) (Tinetti et al., 1986). However, robust evidence in support of these tests to accurately identify fall risk is lacking (Gates et al., 2008; König et al., 2014; Oliver et al., 2008). The reliability, validity, sensitivity, and specificity of these tests in detecting fall risk among older adults can be found in Table 1.1 below.

Table 1.1. Reliability, validity, sensitivity, and specificity of commonly used and recommended clinical mobility tests to assess fall risk among older adults

Test	Reliability	Validity	Sensitivity	Specificity
TUGT	Inter-rater reliability ICC = 0.98 (Shumway-Cook et al., 2000)	Spearman rank correlation = 0.71 - 0.9 (Sebastião et al., 2016)	35.8% (Palumbo et al., 2019)	84% (Palumbo et al., 2019)
BBS	Inter-rater reliability ICC = 0.95 (Major et al., 2013)	Internal consistency Cronbach's alpha = 0.83 (Major et al., 2013)	53% (Bogle Thorbahn et al., 1996).	96% (Bogle Thorbahn et al., 1996).
POMA	Inter-rater reliability ICC = .88-.97 (Sterke et al., 2010)	Internal consistency Cronbach's alpha = 0.94 (Moulodi et al., 2020)	0.27% to 0.76% (Gates et al., 2008)	0.53% to 0.83% (Gates et al., 2008)

A variety of other clinical mobility screening tests exist; however recent systematic reviews assessing the accuracy of various screening tools for fall risk independently concluded that insufficient evidence exists for any singular test to have the specificity and sensitivity needed to correctly identify fall risk (Gates et al., 2008; Lusardi et al., 2017). Further, many of these clinical tests, including the BBS and POMA, suffer from some degree of subjectivity in their measures. Thus, most recommendations include a battery of multiple assessments of gait, mobility, and balance to understand fall risk. Recently, some instrumented versions of this test have shown higher sensitivity and specificity, but no robust method to detect falls from this test alone has been clearly defined (Vervoort et al., 2016; Weiss et al., 2011). The development of a screening tool that is valid, reliable, sensitive and specific for fall risk has been an interest of concern for decades (Martins et al., 2018). As such, there is a need for a composite measure that is easily implementable, as well as accurate and objective to assess fall risk.

Gait Initiation Assessment

In terms of measuring GI, a variety of methods may be implemented. The majority of such include either detection of peak excursion of COP, the entire APA duration, or feature detection to break GI up into its constituent phases. However, there is no standardized approach or understanding for which aspects contain the most important information. In deciding which variables to assess, the literature has a wide scope, measuring muscle activation patterns, kinematic and kinetic variables. Analysis of such large datasets can be highly computationally demanding, and difficult to understand the most salient features for characterizing fallers from non-fallers. This has led to the need for a method to identify and retain all the important variability among gait initiation trials, among large multidimensional datasets (Lu et al., 2017; Winter, 2009; Yiou et al., 2017).

Principal Component Analysis

In understanding differences between populations (i.e., fallers versus non-fallers) in waveform data, it can be difficult to distinguish which features are most important to characterize such differences. Further, in large, multi-dimensional datasets, the computational demand to assess differences between groups can be high. Thus, principal component analysis (PCA) has become a common method of reducing the dimensionality of a dataset while maintaining a large proportion of the variance. Specifically, in studies of specific motor programs from a kinetic and kinematic perspective, a breadth of research has implemented this method to reduce the dataset to the important characteristics without losing any salient information.

Principal component analysis allows large, multidimensional datasets to be reduced to only a few principal components (PCs) necessary to retain a certain desired percentage of the variance from the original dataset, generally about 90-95% (Robertson et al., 2004). PC scores are composed of coefficients measuring the contribution of each PC to the original waveform data, referred to as eigenvalues, as well as the direction of these loading scores, referred to as eigenvectors.

For comparing pathological samples with healthy controls, PCA is a helpful method to understand where differences lie and discriminate between groups. Deluzio et al., (2007) examined knee flexion angles during walking gait among healthy controls compared to individuals with knee osteoarthritis. Waveform data from each group were compared through PCA, and 90% of the original dataset variance was retained through the retention of only three PCs. Further statistical analysis on these PC scores revealed which PC had the highest discriminatory power between the healthy controls and the knee osteoarthritic group.

Specific to gait initiation, Muniz et al., (2012) investigated patients with Parkinson’s Disease who had undergone bilateral subthalamic stimulation, compared to those who had not. Only the onset of the APA was investigated, including the vertical GRF shift from onset to toe-off on the initial swing leg. Principal component analysis was used to generate PCs for both the healthy and the PD groups to allow for comparison. Principal component analysis provides a robust method of dataset reduction while maintaining the important directions of variance critical to a holistic understanding of the dataset. PCA takes out any subjectivity associated with feature detection. As the computed PCs are all independent of each other, they reduce dataset redundancy and ensure only the most important parts of the dataset are used for further statistical analyses.

Chapter 2 — Detecting differences in gait initiation between older adult fallers and non-fallers through time-series principal component analysis (PCA)

Introduction

Falls are a growing problem among the increasingly large aging population in North America (Lusardi et al., 2017; Murphy et al., 2003; Scott et al., 2010). It is estimated that nearly one in every three adults aged 60+ will experience a fall annually, often resulting in decreased mobility and independence due to injury and fear of falling again (Lusardi et al., 2017; Murphy et al., 2003; Scott et al., 2010). Fall risk in older adults is thought to arise, in part, from the age-related deterioration of postural control which requires the complex integration and coordination of sensory and neuromuscular systems (Aleksandra et al., 2006). While there are currently few objective, clinically feasible and robust fall-risk detection methods, changes in functional tasks such as walking, have provided important clinical insight (Boulgarides et al., 2003; Lusardi et al., 2017). For example, in steady-state walking gait, there are well-described differences between fallers and non-fallers, such as shorter stride length, decrease in gait velocity and increased double-support time (Ambrose et al., 2013; Barak et al., 2006; Van Schooten et al., 2016; Verghese et al., 2009). While differences in steady-state walking metrics appear to be valuable indicators of fall risk it has been demonstrated that falls often occur during short, transitional tasks such as gait initiation (GI) (Callisaya et al., 2016; Robinovitch et al., 2013; Suzuki et al., 2004; Tinetti et al., 1995).

Gait initiation is the transitional phase between static balance in an upright position and the start of walking (Chang & Krebs, 1999) and encompasses anticipatory postural adjustments (APA) and the propulsive joint forces necessary to generate the first step of walking gait (Winter, 2009). This task is often examined using force plates to assess the ground reaction forces (GRF) and movement of the centre of pressure (COP) that accompanies the GI phases. While there is a robust dataset of continuous time-series kinematics and kinetics collected during GI, simple statistical assessments of discrete variables are often enough to determine differences between populations. For example, GI differences have been observed between fallers and non-fallers such as a longer overall weight transfer and forward progression phases in GI (Callisaya et al., 2016), decrease in propulsive force generation, as well as shorter and more variable first step lengths for fallers as compared to non-fallers (Ellmers et al., 2020; Mbourou et al., 2003). While these simple assessments of GI data are valuable, there have been more complex approaches to GI data analysis that have yielded important findings in other clinical populations. For example, Muniz et al (2012) used a continuous time-series waveform PCA approach to determine differences in GI kinematics and kinetics between PD and healthy subjects (Muniz et al 2012). Specifically, using a PCA approach to analyze time-series data, from COP kinematics and GRF kinetics, enabled the researchers to reduce the dataset to maintain only those differences across groups that retained the maximal variance and contain information about the magnitude, timing and pattern of movement and forces. The analysis showed a marked reduction in APA amplitude among the PD subjects, suggesting a functionally reduced capacity for GRF generation during GI (Muniz, 2012). Therefore, PCA analysis of time-series data provides a powerful technique to examine changes in magnitude, pattern and timing of the data that may not be observed by discrete comparisons alone and may enhance functional task comparisons within and between clinical populations. However, there has yet to be a detailed assessment of GI kinematics and kinetics using time-series PCA analysis, alongside discrete variable comparisons, to evaluate differences between faller and non-faller cohorts. Therefore, this study aims to characterize differences between fallers and non-fallers by using time-series PCA of GI kinematic and kinetics, alongside discrete variable comparison, to yield important functional differences during gait initiation that can be used to support fall risk assessment and interventions.

Methods

Research Design and Participant

In this quasi-experimental, cross-sectional design, participants were classified into fallers (*N*= 28, 21F, 7M) and non-fallers (*N*=28, 16F, 12M) based on self-reported fall status for one or more falls occurring within the 12 months before participation in the study. Falls were defined as “an event which resulted in a person coming to rest unintentionally on the ground or other lower level, not as the result of a major intrinsic event or an overwhelming hazard” (Lord et al., 1994), and the circumstances of each reported fall were reviewed to match this definition. The researchers were blind to participant fall status until after the assessment. All participants were aged 65 years or older, community-dwelling, and able to walk one city block (approximately 100 meters) unassisted. Further, all participants included in this study were fluent in English and obtained a Mini-Mental State Examination (MMSE) score greater than 24 (Folstein et al., 1975). Exclusion criteria included a diagnosis of dementia, recent major illness or neurological, sensory, or mobility impairment, or musculoskeletal disorders or injuries.

Recruitment

A sample of 56 community-dwelling older adults (77.8 ± 5 years) were recruited through purposive sampling in the Greater Victoria area. The required sample size was calculated in G*power with a power of 0.80 and a literature-derived effect size of 0.55 based on overall gait initiation time for fallers vs. non-fallers (Callisaya et al., 2016). Participants received compensation for parking fees for the assessment. Participants provided written informed consent to procedures approved by the Human Research Ethics Board at the University of Victoria.

Data collection and Procedures

All subjects were instructed to wear comfortable footwear to the assessment. Height, weight, foot length, and MMSE scores were recorded upon assessment, followed by tests of GI. Participants began standing with both feet on one force plate, and when instructed by a verbal cue of “whenever you’re ready”, began walking at a self-selected pace to the end of a 6.4 m walkway, as illustrated in Figure 2.1 below. Participants chose which foot to begin walking on and were encouraged to walk normally. Three practice trials were completed by each participant to allow familiarization with the research protocol, followed by five recorded GI trials.

Instruments

Two 40 x 60 cm triaxial Kistler force plates (Model #:9286AA) Kistler Instrument Corp., Winterthur, Switzerland) were arranged in sequence and the distance between them was adjusted during the familiarization protocol so that the participant’s first step landed on the second force plate. The force plates were used to capture ground reaction forces (GRF) in the mediolateral (ML), anteroposterior (AP) and vertical directions. In the ML plane, left was defined as negative and right as positive. In the AP plane, posterior movement was defined as negative and anterior movement as positive. Data were collected at a sampling rate of 1024 Hz.

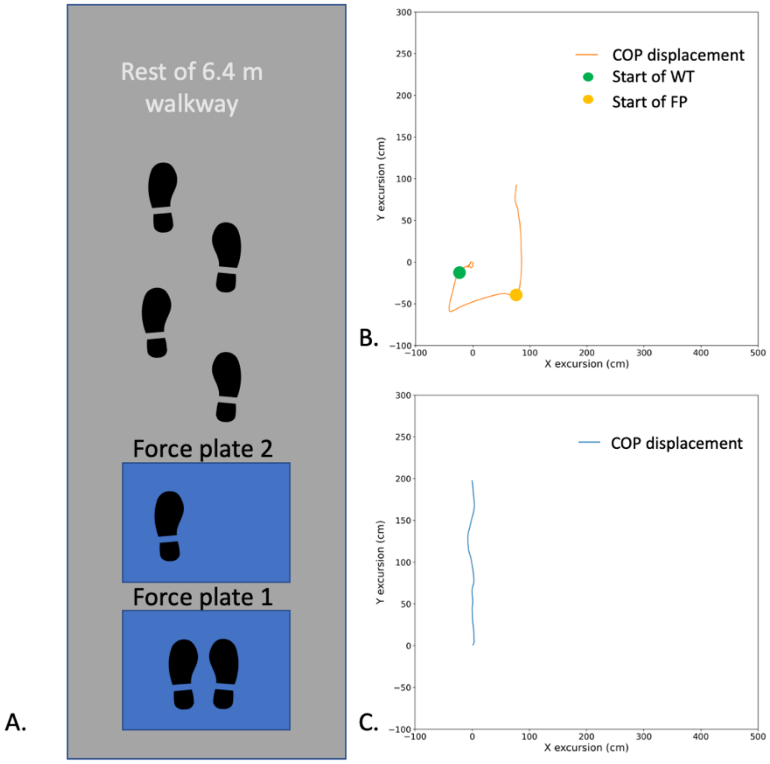


Figure 2.1. Data collection procedure.
A) Experimental setup for gait initiation trials. Participants start with both feet in a comfortable stance on force plate one and begin walking at a self-selected pace.
B) Center of pressure traces from force plate one in an example gait initiation trial. The green dot on force plate one indicates the programmatic detection of the beginning of weight transfer, after premovement, and the yellow dot indicates the programmatic detection of the beginning of forward progression.
C) Center of pressure traces from force plate two in an example gait initiation trial.

Data Analysis

Participant characteristics

Descriptive statistics and between group differences based on independent *t*-tests for all participant characteristics including age, sex, height, foot length and MMSE scores between the faller and non-faller groups were computed. The alpha level for statistical significance was set at 0.05 for all analyses.

Signal Processing

Signal processing was done in custom-written LabVIEW software (LabVIEW 2019, National Instruments) and Python script (Python Version 3.8, Python Programming Software). Data were filtered using a 30 Hz low pass, second-order, double Butterworth filter (Muniz & Nadal, 2009). The time-series data were then segmented into different kinetic and kinematic components including forces in the ML, AP, and vertical axes, as well as the area of excursion in the ML and AP directions. Next, GI trials were broken up into four distinct phases: (1) Premovement (2) Weight Transfer, (3) Forward Progression, and (4) First Step.

(1) Premovement was defined as the period 300 milliseconds (ms) before the onset of weight transfer.

(2) Weight transfer, when both feet are on the ground pre-swing, was detected from the differences between adjacent samples of the ML COP displacement signal. The first value exceeding a threshold based on the mean of this signal plus 2x the standard deviation of the signal was defined as the onset of weight transfer, until the beginning of forward progression. This was visually inspected for accuracy and adjusted where necessary.

(3) Forward progression, when all the bodyweight was on the stance limb, was detected from AP COP displacement signal. This signal was reversed, and we determined the first valley that crossed a threshold of the mean of the last second as the onset of forward progression, and the end of weight transfer. This was visually inspected for accuracy and adjusted where necessary. The temporal duration of forward progression, known as the swing time of the initial swing foot, extends until the start of the first step is detected on the second force plate.

(4) First step, from the initial contact of the swing foot to toe-off, was defined as the entire signal detected by the second force plate once the force in the vertical axis exceeded a threshold of $< 1\%$ of the maximum signal value.

Data were normalized so that all subjects apparently stepped with their left foot first. For those that stepped with their right foot first, the ML signals from force plate one and force plate two were negated. To correct for the location of foot placement on each force plate, the initial ML position was subtracted to remove the bias. To correct for the location of foot placement on each force plate, the initial ML position was subtracted to remove the bias. This only applies to normalization in the ML direction. For force plate two, the orientation of the foot trajectory was normalized using linear interpolation between initial contact and toe-off and subtracting this from the original signal to remove bias. This method means that the AP movements are now relative to the AP axis of the foot and ML movements are relative to the ML axis of the foot. All phases of the kinematics AP signal from force plate one were zeroed to create a consistent starting foot location, except for forward progression, which underwent the same bias removal technique used for foot placement.

Kinematic data

The signal was aligned such that data from the ML portion of the signal represented the first 100 samples, followed by the AP portion of the signal as the next 100 samples for each phase. Based on this, $p = 200$ for each phase, creating a 280 x 200 (56 participants x 5 trials each = 280 rows) matrix for each phase. The data were then resampled to 280 time points to create a symmetrical 280 x 280 matrix for each phase, shown in equation 1 below.

$$X = \begin{matrix} x_{11} & x_{12} & \cdots & x_{1p} \\ x_{21} & x_{22} & \cdots & x_{2p} \\ \vdots & \vdots & \ddots & \vdots \\ x_{n1} & x_{n2} & \cdots & x_{np} \end{matrix} \tag{1}$$

Kinetic data

All force data were normalized to the measured bodyweight and displayed as % bodyweight (%BW). The same procedure as described for kinematic data was implemented for kinetic data to create matrix A, where each phase was composed of 100 time points for each signal (ML, AP, and vertical forces), thus $p = 300$ for each phase, creating a 280 x 300 matrix. Again, this data was resampled to 280 time points to create a symmetric matrix, X.

Principal Component Analysis

Principal component analysis (PCA) was applied to each phase of the kinetic and kinematic data matrices. Separately for both the kinematic and kinetic data, the mean was removed from matrix X, and a covariance matrix was calculated. The corresponding covariance matrix (*S*) was calculated based on the original matrix, given by equation 2:

$$S = \frac{\sum_{k=1}^{n}(x_{ki} - \bar{x}_i)^2}{n-1} \tag{2}$$

Where *n* = number of rows (280 trials), and *i* = columns (Robertson et al., 2004). The diagonal elements of this matrix represent the variance at each instant based on the above equation (2). The principal components (PC) of this data were extracted from matrix *S*, determined based on the eigenvalues given by the columns of the solution to the linear system. The covariance matrix *S* was transformed into the principal component's covariance matrix *C* through orthogonal decomposition given by equation 3:

$$C = U^t S U \tag{3}$$

Where matrix *U* is an orthogonal transformation matrix realigning the data to a new coordinate system. The columns (time points) of matrix *U* are the eigenvectors of *S* (Robertson et al., 2004). The eigenvector explaining the largest percentage of signal variance is the extracted and retained as the first PC, followed by the second, and so on, until 90% of the variance was explained (Kobayashi et al., 2014). Finally, individual loading scores for each subject by PC are computed by first taking the product of the original scores at each time point for each individual GI (*p* (*n* = 280) and the computed eigenvectors used to calculate each PC, given by the values on the right side of equation 4. Then, these scores for each *p* were summed to create individual loading scores for each point in the time-series and sorted by fall status for further analysis, as shown in equation 4 (*SAS-STAT User's Guide*, 1988)

(4) $$z_1 = b_{11}(x_1) + b_{12}(x_2) + \cdots\ b_{1p}(x_p)$$

Where:

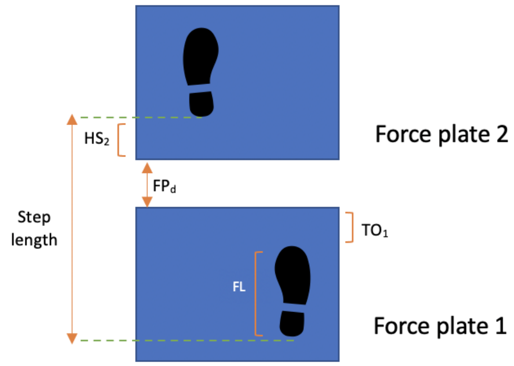
z_1 = the subject's score on PC 1 for trial 1

b_{1p} = the eigenvector (or weight) for observed time point p, as used in calculating PC 1

x_p = the subject's score at time point p for each individual GI

Temporospatial metrics

Weight transfer time, forward progression time, first step time, ground contact time, first step length, first step width, and ML excursion and AP excursion during weight transfer, forward progression and first step were computed and included in between groups analysis. All temporal variables were measured in seconds, and all spatial variables were measured in cm. Ground contact time was calculated as the time in contact with the second force plate. The first step length was measured from toe-off on the initial stance foot, to heel-strike on the initial swing foot, summed with foot length and force plate distance to determine the complete heel-to-heel distance, shown in Figure 2.2 below.



Where:

TO$_1$ = Final Y position at toe off on force plate 1
HS$_2$ = Initial Y position at heel strike on force plate 2
FP$_d$ = distance between force plate 1 and 2
FL = Foot length

$$Step\ length = FL + TO_1 + FP_d + HS_2$$

Figure 2.2. First step length calculations

The first step width was calculated as the difference between the final X position on force plate one, and the initial X position on force plate two in mm. Excursion in the ML direction was calculated as the total displacement in the ML direction throughout the weight transfer phase, measured from the COP path. This was also calculated for the AP direction.

Statistical Analysis of Individual Loading Scores

A one-way, multivariate analysis of covariance (MANCOVA) was performed on the retained individual loading scores for each PC by phase for each trial of GI, with fall status as the fixed factor in SPSS (SPSS Inc. version 20.0, 2016, Chicago, IL). Weight transfer time, forward progression time, first step time, ground contact time, step width and step length (see Table 3) were included as covariates. Where indicated, post-hoc comparisons with Bonferroni corrections were used to assess group differences between fallers and non-fallers. To further understand what part of the signal these differences represent, the PC was graphed against the original signal of averages for fallers and non-fallers. Between groups, independent *t*-tests were computed on all temporospatial variables described above, as well as ML and AP excursion during weight transfer, forward progression and first step, by fall status, to assess group differences of temporospatial metrics of GI.

Results

Participant characteristics

All participant characteristics are presented in Table 2.1 below. No differences were found between groups in terms of anthropometric variables or MMSE scores.

Table 2.1. Participant characteristics by group at baseline.
Data is represented by mean (SD); p-value based on alpha < 0.05.

	Age (years)	Height (m)	Weight (kg)	BMI (kg/m^2)	MMSE score (/30)	Foot length (cm)
Non-fallers (*N*=28)	78.11 (5.32)	1.66 (0.1)	75.36 (14.55)	27.2 (3.81)	28.61 (1.03)	26.22 (3.41)
Fallers (*N*=28)	77.54 (4.91)	1.66 (0.1)	71.29 (13.32)	26.37 (4.47)	28.77 (0.1)	26.36 (2.32)
p-value	0.68	0.96	0.28	0.09	0.58	0.86

Temporospatial metrics

The results of the independent, two-tailed *t*-tests are indicated in Table 2.2 below. Differences were observed between fallers and non-fallers for first step length ($p < 0.01$) and ML excursion during weight transfer ($p = 0.008$). Fallers had a demonstrably shorter first step, compared to the non-falling cohort. Fallers also had a greater range of excursion in the ML direction during the weight transfer phase.

Table 2.2. Temporospatial variables by group.
Data is represented by mean (SD), and maximum, minimum by group; p-value based on alpha < 0.05.

	Non-fallers (*N*=28)	Fallers (*N*=28)	Total (*N*=56)	*p*-value
Weight transfer time (s)	0.97 (0.251)	0.991 (0.184)	0.981 (0.22)	0.222
Max, min (s)	1.77, 0.238	1.538, 0.296		
Forward progression time (s)	0.492 (0.085)	0.476 (0.079)	0.484 (0.082)	0.074
Max, min (s)	0.711, 0.21	0.663, 0.286		
First step time (s)	0.568 (0.096)	0.551 (0.084)	0.56 (0.09)	0.119
Max, min (s)	0.916, 0.384	0.747, 0.351		
Ground contact time (s)	0.684 (0.081)	0.689 (0.077)	0.686 (0.079)	0.594
Max, min (s)	0.979, 0.533	0.916, 0.509		
First step length (cm)	67.444 (9.135)	61.309 (10.937)	64.377 (10.517)	**< 0.000***
Max, min (s)	93.934, 45.982	86.734, 40.903		
First step width (cm)	16.449 (4.774)	16.578 (5.486)	16.514 (5.134)	0.924
Max, min (cm)	30.015, 0.053	31.194, 4.83		
WT ML excursion (cm)	11.579 (3.337)	12.917 (3.988)	12.332 (3.711)	**0.008***
Max, min (cm)	21.747, 6.039	22.092, 5.864		
WT AP excursion (cm)	4.720 (1.475)	4.995 (1.517)	4.870 (1.513)	0.169
Max, min (cm)	10.064, 1.581	8.287, 1.966		
FP ML excursion (cm)	2.262 (1.509)	2.253 (1.227)	2.277 (1.406)	0.782
Max, min (cm)	7.728, 0.531	8.551, 0.585		
FP AP excursion (cm)	16.841 (4.476)	18.130 (5.045)	17.567 (4.944)	0.057
Max, min (cm)	28.726, 9.396	28.244, 8.409		
FS ML excursion (cm)	2.084 (1.233)	2.262 (1.626)	2.182 (1.467)	0.357
Max, min (cm)	9.163, 0.469	7.379, 0.653		
FS AP excursion (cm)	22.292 (3.739)	22.801 (3.264)	22.633 (3.513)	0.420
Max, min (cm)	29.909, 12.213	29.505, 12.674		

* and bold indicates significance at alpha < 0.05 between cohorts. WT = weight transfer, FP = forward progression, FS = first step

Figure 2.3. below shows the average COP displacements for fallers and non-fallers throughout gait initiation.

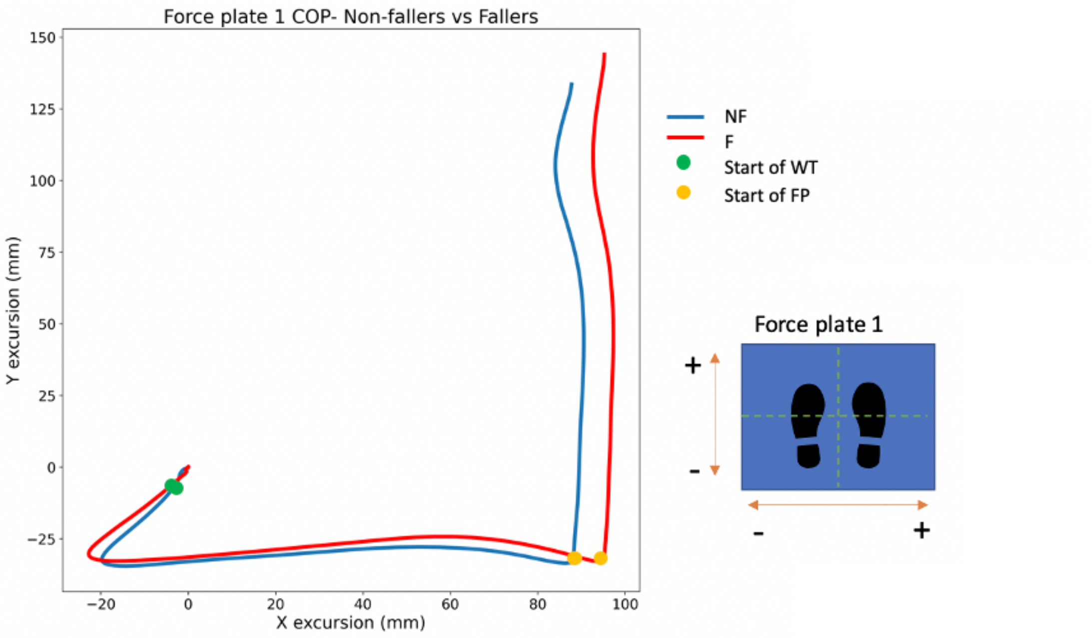


Figure 2.3. Average COP trace for the fallers and non-fallers on force plate 1. NF = non-faller, F = faller, PM = premovement, WT = weight transfer, FP = forward progression.

Kinematic data

To achieve at least 90% variance explained, six PCs were retained for weight transfer (92.43%), two PCs for forward progression (90.08%) and four PCs for first step (90.77%), illustrated in Table 2.3. The individual loading scores based on each retained PC were computed and included in the MANCOVA analysis as the dependant variables.

Table 2.3. Principal component models for kinematic data based on the 90% variance.

	Premovement	Weight transfer	Forward progression	First step
Variation explained (%)	90.29	92.43	90.08	90.77
Number of PCs	2	6	2	4

Based on this, a 2 (fall status) x 14 (individual loading scores for each PC) between-subjects multivariate analysis of covariance (MANCOVA) was performed on the 14 dependant variables, after controlling for the temporospatial covariates described above. The independent variable was fall status (0 = non-faller, 1 = faller). There was a difference between the fallers and non-fallers on the combined dependant variables of the individual loading scores after controlling for the covariates listed above, F (14, 259) = 2.771, $p < 0.01$, Wilks $\Lambda = .869$ partial $\eta^2 = .133$. Post-hoc comparisons with Bonferroni corrections indicated that the main effects of fall status were significant on the individual loading scores for PC 1 of premovement ($p = 0.021$), PC 4 of weight transfer ($p = 0.036$) and PC 1 of forward progression ($p = 0.019$), shown in Table 2.4 below.

Table 2.4. Results of MANCOVA on individual loading scores for each PC of kinematic data by fall status for each phase.

Individual loading scores on PC by phase	**Mean difference**	**Standard error**	**Significance**	**95% Confidence Interval**	**NF, F* mean loading on PC**
Premovement					
PC 1	0.725	0.312	**0.021***	[0.111, 1.339]	**Negative, positive**
PC 2	-0.234	0.375	0.534	[-0.972, 0.505]	-
Weight Transfer					
PC 1	-5.297	3.617	0.144	[-12.417, 1.823]	-
PC 2	-3.748	2.342	0.111	[-8.359, 0.863]	-
PC 3	-3.740	2.403	0.121	[-8.472, 0.991]	-
PC 4	-7.792	3.694	**0.036***	[-15.064, -0.52]	**Positive, negative**
PC 5	-3.300	2.617	0.208	[-8.453, 1.853]	-
PC 6	-0.349	3.393	0.918	[-7.028, 6.33]	-
Forward Progression					
PC 1	6.793	2.881	**0.019***	[1.122, 12.465]	**Negative, positive**
PC 2	4.961	2.755	0.073	[-0.462, 10.384]	-
First Step					
PC 1	-4.249	2.677	0.114	[-9.518, 1.021]	-
PC 2	2.154	4.514	0.634	[-6.733, 11.042]	-
PC 3	0.963	4.596	0.834	[-8.086, 10.013]	-
PC 4	**1.435**	**1.666**	**0.390**	**[-1.846, 4.716]**	**-**

* and bold indicates significance at alpha < 0.05, NF = non-faller, F = faller

For premovement, it was revealed that PC 1 explained 57.06% of the signal variance. The mean signal of the non-faller time-series weighed negatively on this PC, and the fallers weighed positively. On average, it was observed that fallers had less lateral COP displacement toward the initial stepping foot, compared to the non-fallers. Moreover, fallers had less posterior displacement leading into weight transfer, compared to the non-fallers. The differences for both the ML and AP components are shown in Figures 2.4 A and B below.

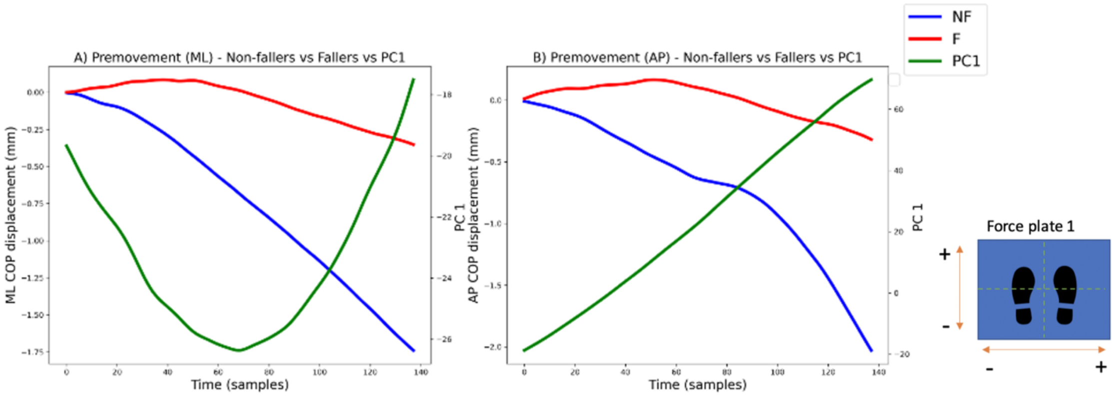


Figure 2.4. Premovement phase for mediolateral (A) and anteroposterior (B) COP displacements (mm), shown as the average of non-fallers, fallers, and PC 1 loading vector.
The force plate diagram illustrates the direction of negative and positive movements during the phase.

For weight transfer, PC 4 explained 7.11% of the signal variance. The mean signal of the non-faller time-series weighed positively on this PC, and the fallers weighed negatively. The shape and sign of the PC would be scaled to the magnitude of either the faller or the non-faller cohort. This signifies the general trend of differences between these two cohorts, as seen by the green line in the Figures below, currently in the positive configuration. Specifically, in the ML axis, fallers move further towards the initial stance limb at the end of the phase, compared to the non-fallers. In the AP axis, fallers have less posterior displacement, compared to the non-fallers. Shown in Figure 2.5 A and B below.

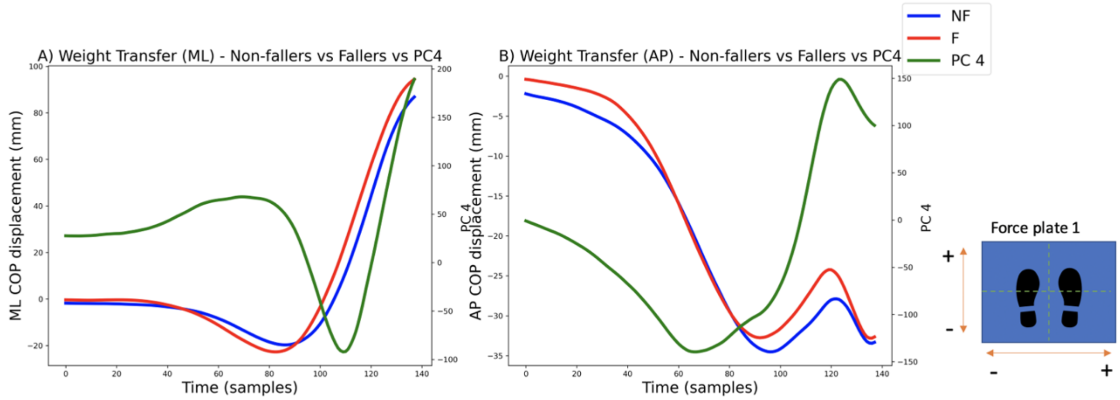


Figure 2.5. Weight transfer phase for mediolateral (A) and anteroposterior (B) COP displacements (mm), shown as the average of non-fallers, fallers, and PC 4 loading vector. The force plate diagram illustrates the direction of negative and positive movements during the phase.

For forward progression, PC 1 explained 63.70% of the signal variance. The mean signal of the non-faller time-series weighed negatively on this PC, and the fallers weighed positively. In the ML axis, the fallers maintain a more lateral COP position on the initial stance limb throughout the phase, compared to the non-fallers. In the AP axis, the fallers have slightly greater anterior displacement, shown in Figure 2.6 below.

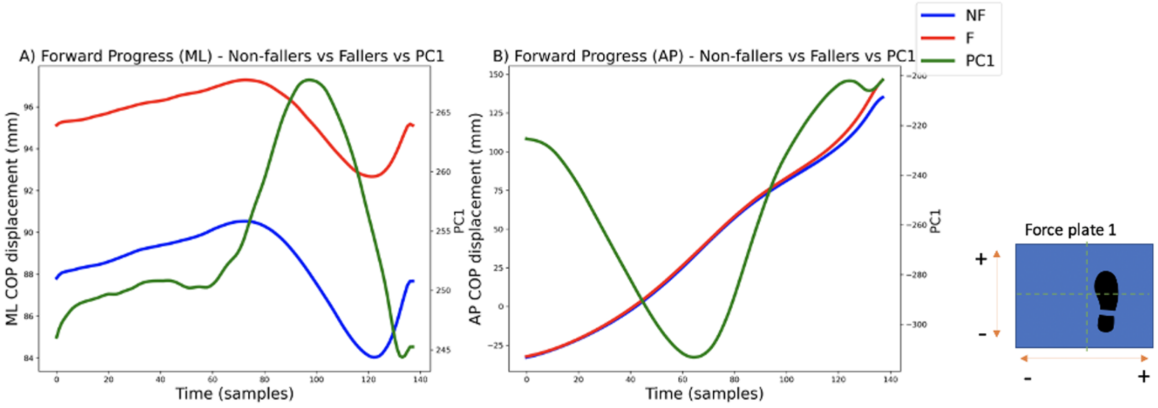


Figure 2.6. Forward progress phase for mediolateral (A) and anteroposterior (B) COP displacements (mm), shown as the average of non-fallers, fallers, and PC 1 loading vector. The force plate diagram illustrates the direction of negative and positive movements during the phase.

No statistical differences were found between fallers and non-fallers in kinematic variables for first step (Figure 2.7).

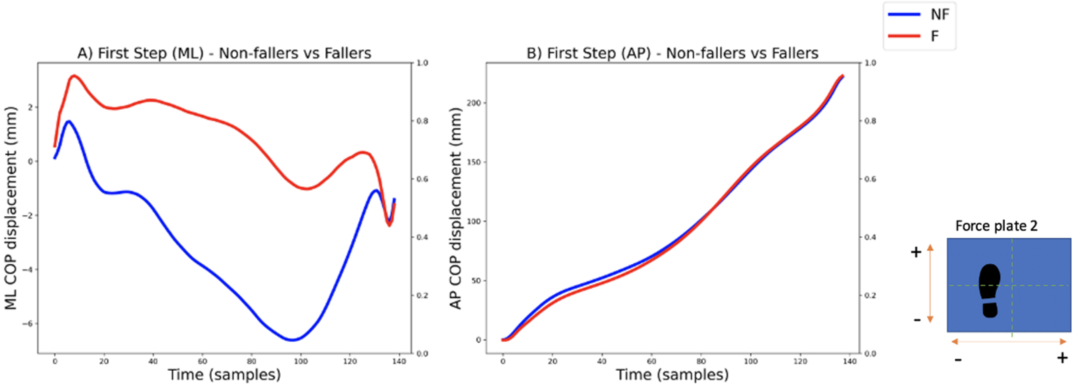


Figure 2.7. First step phase for mediolateral (A) and anteroposterior (B) COP displacements, shown as the average of non-fallers, fallers.
The force plate diagram illustrates the direction of negative and positive movements during the phase.

Kinetic data

To retain 90% of the variance of the original data set, individual loading scores for the first eight PCs were retained for premovement (91.26%), eleven PCs for weight transfer (90.26%), five PCs for forward progression (92.79%), and six PCs for first step (91.60%) illustrated in Table 2.5.

Table 2.5. Principal component models for kinetic data based on the 90% variance.

	Premovement	Weight transfer	Forward progression	First step
Variation explained (%)	91.26	90.26	92.79	91.60
Number of PCs	8	11	5	6

Based on this, a 2 (fall status) x 30 (individual loading scores for each PC) between-subjects MANCOVA was performed on the 30 dependant variables, after controlling for the temporospatial covariates described above. The independent variable was fall status (0 = non-faller, 1 = faller). There was a difference between the fallers and non-fallers on the combined dependant variables of individual loading scores on each PC after controlling for covariates listed above, F (30, 243) = 1.610, *p* = 0.028, Wilks Λ = .834 partial η^2= .198. Post-hoc comparisons with Bonferroni corrections indicated differences exist between groups, shown in Table 2.6 below. The main effects of fall status were significant on the individual loading scores for PC 4 of weight transfer (*p* = 0.007), and PC 2 of first step (*p* = 0.009).

Table 2.6. Results of MANCOVA on individual loading scores for each PC of kinetic data by fall status for each phase.

Individual loading scores on PC by phase	**Mean difference**	**Standard error**	**Significance**	**95% Confidence Interval**	**NF, F* mean loading on PC**
Premovement					
PC 1	0.000	0.000	0.709	[-0.001, 0.001]	-
PC 2	-0.001	0.000	0.064	[-0.001, 0.00004]	-
PC 3	0.000	0.000	0.436	[-0.001, 0.001]	-
PC 4	0.001	0.001	0.224	[0, 0.002]	-
PC 5	0.000	0.000	0.802	[-0.001, 0.001]	-
PC 6	0.000	0.000	0.539	[-0.001, 0.001]	-
PC 7	0.000	0.000	0.913	[-0.001, 0.001]	-
PC 8	-0.001	0.001	0.232	[-0.002, 0.001]	-
Weight Transfer					
PC 1	0.002	0.002	0.495	[-0.003, 0.006]	-
PC 2	-0.004	0.002	0.091	[-0.009, 0.001]	-
PC 3	-0.001	0.004	0.906	[-0.009, 0.008]	-
PC 4	.014	0.005	**0.007***	[0.004, 0.025]	**Negative, positive**
PC 5	0.003	0.004	0.330	[-0.004, 0.01]	-
PC 6	-0.003	0.003	0.272	[-0.009, 0.003]	-
PC 7	0.003	0.004	0.410	[-0.004, 0.011]	-
PC 8	0.007	0.006	0.225	[-0.004, 0.018]	-
PC 9	0.002	0.005	0.653	[-0.007, 0.012]	-
PC 10	-0.001	0.003	0.748	[-0.008, 0.006]	-
PC 11	-0.007	0.003	0.052	[-0.014, 0.00006]	-
Forward Progression					
PC 1	0.000	0.005	0.974	[-0.01, 0.01]	-
PC 2	0.009	0.008	0.259	[-0.007, 0.025]	-
PC 3	0.002	0.006	0.693	[-0.009, 0.014]	-
PC 4	-0.009	0.005	0.076	[-0.018, 0.001]	-
PC 5	0.004	0.003	0.307	[-0.003, 0.01]	-
First Step					
PC 1	0.009	0.008	0.291	[-0.008, 0.025]	-
PC 2	0.020	0.012	**0.009***	[-0.003, 0.043]	**Negative, positive**
PC 3	-.025	0.012	0.035	[-0.049, -0.002]	-
PC 4	-0.001	0.013	0.923	[-0.027, 0.025]	-
PC 5	-0.001	0.014	0.926	[-0.028, 0.026]	-
PC 6	0.003	0.007	0.715	[-0.011, 0.016]	-

*and bold indicates significance at alpha < 0.05, NF = non-faller, F = faller

No statistical difference was found between groups for kinetic variables of premovement (Figure 2.8).

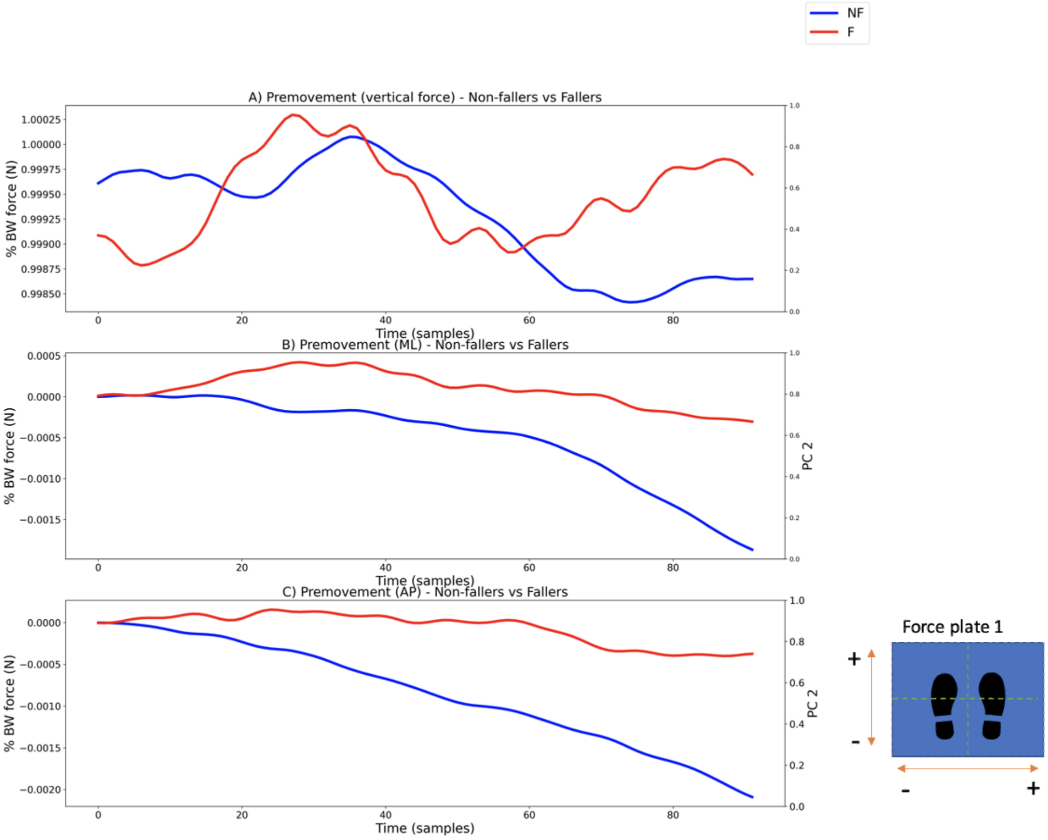


Figure 2.8. Premovement phase for kinetic vertical (A), mediolateral (B), and anteroposterior (AP) forces, shown as the average of non-fallers, fallers.
The force plate diagram illustrates the direction of negative and positive movements during the phase.

Of the weight transfer phase of kinetic variables, PC 4 explained 6.22% of the variance of the phase. The mean signal of the non-faller time-series weighed negatively on this PC, and the fallers weighed positively. The faller group had a faster rate of force development posteriorly, followed by larger anteriorly directed ground reaction forces, compared to the non-fallers. Within the ML portion of the signal, fallers had less leftward lateral force pushing off from the initial swing leg to transfer weight to the initial stance leg. Fallers also demonstrate lower vertical forces toward the end of the phase. All three components of the signal for non-fallers, fallers and PC 4 are depicted in Figure 2.9 below.

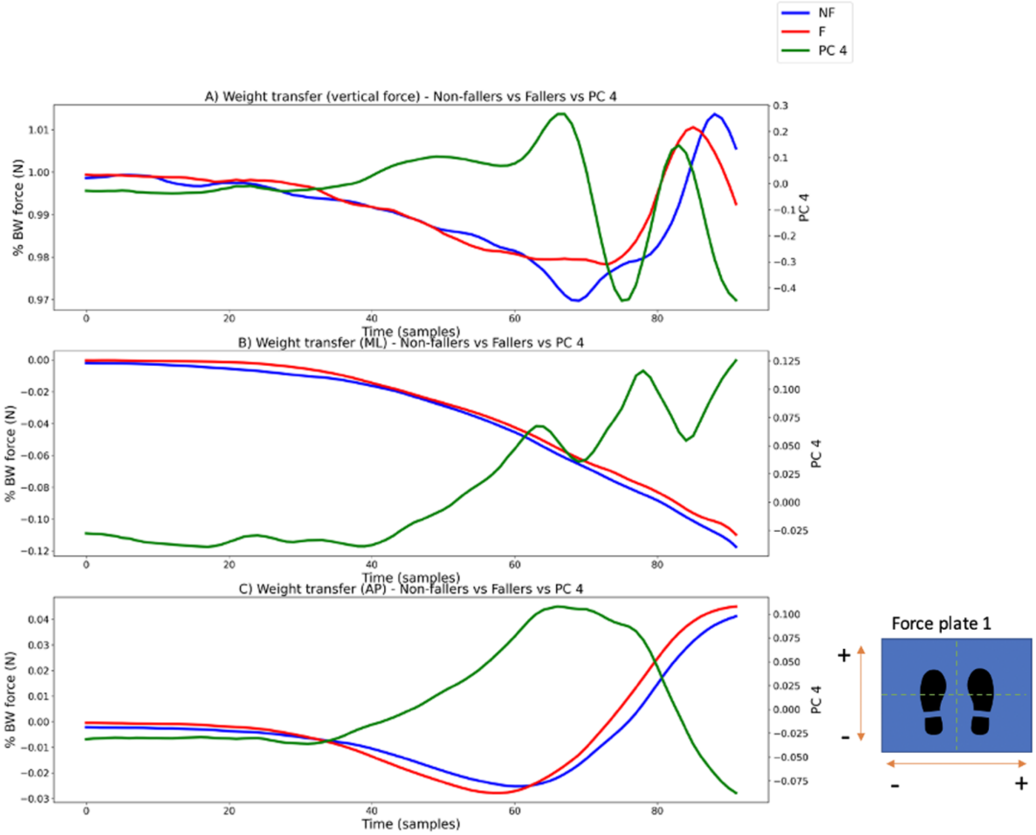


Figure 2.9. Weight transfer phase for kinetic vertical (A), mediolateral (B), and anteroposterior (AP) forces, shown as the average of non-fallers, fallers, and PC 4 loading vector.
The force plate diagram illustrates the direction of negative and positive movements during the phase.

No statistical differences were found for forward progression (Figure 2.10.) between groups in terms of kinetic variables.

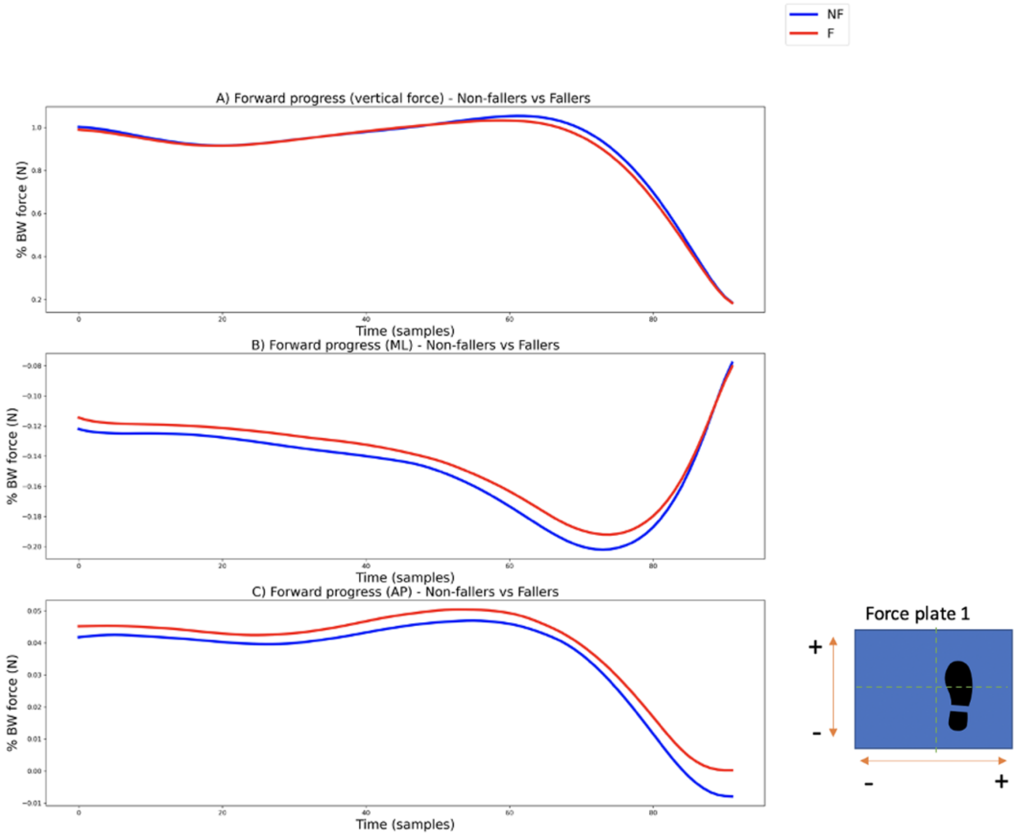


Figure 2.10. Forward progress phase for kinetic vertical (A), mediolateral (B), and anteroposterior (AP) forces, shown as the average of non-fallers, fallers.
The force plate diagram illustrates the direction of negative and positive movements during the phase.

Last, PC 2 of first step kinetic data explained 22.48% of the variance of the first step phase. The mean signal of the non-faller time-series weighed negatively on this PC, and the fallers weighed positively. This signal is characterized by weight acceptance on the second force plate. In the ML direction, among the fallers, there is a lower lateral force component toward the right in the weight acceptance of the first step of the left foot. Antero-posteriorly, fallers have larger anteriorly directed force compared to the non-fallers throughout heel strike, to mid-stance and weight acceptance of the first step. As they roll through the first step towards toe-off, the force anteriorly remains higher among fallers. In the vertical direction, fallers exert less force at heel strike to mid-stance, and again through toe-off into the subsequent step compared to non-fallers (Figure 2.11).

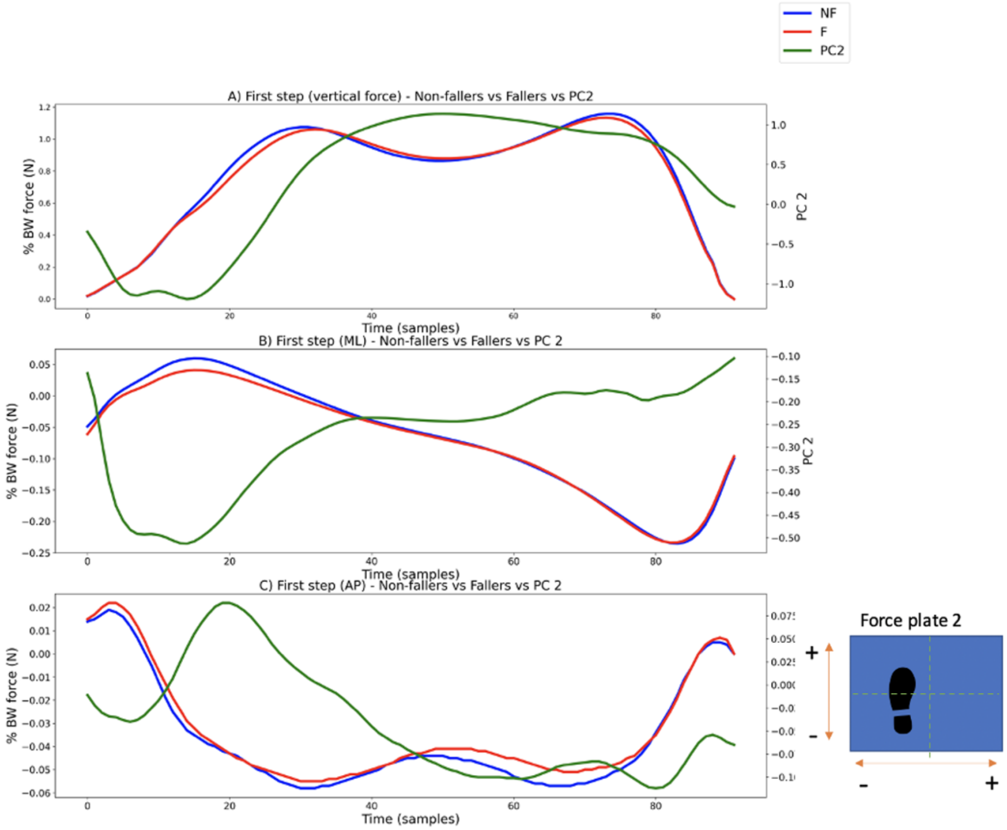


Figure 2.11. First step phase for kinetic vertical (A), mediolateral (B), and anteroposterior (AP) forces, shown as the average of non-fallers, fallers, and PC 2 loading vector.
The force plate diagram illustrates the direction of negative and positive movements during the phase.

Discussion

In this study, time-series principal component analysis (PCA) alongside discrete variable comparison of gait initiation (GI) revealed important differences concerning the magnitude, timing, and pattern of kinematic and kinetic differences between faller and non-faller cohorts that may add to the understanding of fall-risk related balance control strategies. The first finding indicates that during weight transfer and the beginning of forward progression, fallers in this study have a different range of mediolateral (ML) movement, evident in both discrete and PCA results. While having a greater range of ML movement during weight transfer, fallers also have lower force towards the stepping limb. This finding could highlight important differences between fallers and non-fallers with respect to controlling ML stability. The second finding, during weight transfer, shows that fallers have a greater anteroposterior (AP) range of movement of the center of pressure (COP) with a different pattern of AP forces. This could be related to strategies to control forward progression. The third finding indicates that during first step, fallers have a more gradual rise in vertical force, a lower medial force and an increased anterior force during weight acceptance followed by a lower posterior force during heel rise. There is also an observed smaller step length, and these findings together demonstrate an altered approach to forward progression in fallers. Taken together, time-series PCA alongside discrete variable analysis provides a more comprehensive assessment of gait initiation that supports previous findings that fallers may prioritize ML and AP stability over forward progression performance.

The use of time-series PCA for data reduction of kinematic and kinetic variables in this study provided insight on specific differences between fallers and non-fallers during GI that may not be detectable using discrete analysis methods alone. This robust method of data reduction considers all features of the original time-series and results in a condensed set of uncorrelated features that explain maximal variance, removing any subjectivity and overlap in feature extraction (Deluzio & Astephen, 2007). As each principal component (PC) represents a unique feature of the original time-series data, this enables interpretation of each feature by examining the shape of the PC and how each individual's movement pattern loads onto each PC (Deluzio & Astephen, 2007). This allows for the evaluation of entire movement patterns across the complete time-series data and enriches our understanding of the mean discrete data. The PCA of GI in this study specifically allows for characterization of the ML and AP kinematics and kinetics across the time-series which have been identified as key variables in understanding age-related and fall-related indicators. Movement of the COP in both the ML and AP directions are thought to account for different functional aspects of gait initiation. ML movement is considered to relate to lateral stability and AP movement to control forward progression (Brenière et al., 1987; Yiou et al., 2017). While there are conflicting findings with respect to changes during aging, some studies suggest that ML and AP COP displacements decrease with age and are related to a decreased ability to maintain lateral stability and the generation of forward momentum respectively (Mehdizadeh et al., 2021). While our discrete findings for weight transfer only demonstrated differences between the absolute range of ML COP movement, the time-series PCA identified patterns of movement demonstrating a greater range of COP movement by the fallers in both the ML and AP directions during weight transfer. While the ML and AP COP kinematics and kinetics are not entirely independent, they will be discussed separately in relation to their proposed functional purpose during gait initiation.

Centre of pressure ML movement during weight transfer observed in this study support previous findings by Patla et al., (1993) that during gait initiation, older adults move their COP toward their swing limb first. This strategy is considered sub-optimal for overall GI reaction-time performance and not observed as often in young adults but is considered “safer” when concerning balance during this task (Tisserand et al., 2016). In the present study, both fallers and non-fallers have this general strategy, yet the PCA revealed that the fallers tend to move further towards the swing foot and then also further towards the stance foot during weight transfer. This range of movement is corroborated by the discrete findings of a greater ML excursion in the fallers during weight transfer as well as the different ML positioning observed in the PCA during forward progression. Therefore, this suboptimal strategy is likely magnified in this faller cohort as compared to the non-fallers in attempts to be “safer”. Previous studies have found conflicting results between fallers and non-fallers on this kinematic variable which may highlight the value of using time-series PCA to more closely examine biomechanical differences (Lyon & Day, 1997; Tisserand et al., 2016). Interestingly, while Tisserand et al., (2016) observed no differences between fallers and non-fallers in COP ML excursion, they found that the COP stayed lateral to the whole-body center of mass (COM) in relation to the swing foot for a longer time in fallers than in non-fallers in a GI reaction time task. This increase in duration was thought to be a strategy to reduce ML instability through a more efficient torque that propels the COM towards the stance limb and results in a COM that is shifted further towards the stance limb in fallers. While we found opposite results to Tisserand et al., (2016) with a difference observed in the range of ML COP movement without a difference in timing, according to Tisserand et al., (2016) either combination of results could point to a similar goal to reduce ML instability. They hypothesized that the increased range strategy observed in the present study would only be feasible with sufficient muscular strength in the hip adductors/abductors (Tisserand et al., 2016). Thus, it is possible that the present older adult faller cohort may use this strategy to reduce ML instability, over the non-fallers, through manipulation of the COP excursion rather than timing. It, therefore, reasons that either the altered timing or altered excursion strategy could be considered an indicator of fall risk in community-dwelling older adults. The finding of a difference in ML movement by the faller cohort could also suggest a greater base of support in fallers at the onset of GI in our cohort which was not directly measured in our study but would also support the purpose of potentially decreasing ML instability.

Though there was no difference in discrete AP COP movement during weight transfer in the current study, the time-series PCA showed significant differences with fallers having a greater range of AP movement and forces. Brenière et al., (1987) showed that when subjects were instructed to increase walking speed, they did so by increasing the range of AP movement and velocity through a greater posterior movement of the COP to the edge of the base of support. The posterior displacement results in a larger range to generate momentum for forward progression. In the present study, while the range of AP movement was greater in weight transfer in the faller cohort, the fallers had a smaller step length than the non-fallers which could suggest that this strategy may not be related to the generation of forward momentum but potentially used to carefully control forward progression. It has been observed that fear of falling leads to greater postural sway in GI in both the AP and ML directions (Ellmers et al., 2020). While instability is often thought to be the cause of increased postural sway, recent research suggests that an increase in postural sway may in fact be useful to gain and integrate sensory information about the position of the body in relation to the limits of stability (Carpenter et al., 2010; Ellmers et al., 2020; Murnaghan et al., 2011; Riley et al., 1997; Van Emmerik & Van Wegen, 2002). Therefore, this movement strategy by the faller cohort may be a form of exploratory sway to allow for greater movement planning to enhance stability (Rajachandrakumar et al., 2018). This strategy could explain the altered pattern and increased range in AP COP movement during forward progression noted in the time-series PCA data as well as the approaching significant (0.057) difference in the discrete results between the fallers and non-fallers. This could also support a smaller step length and gradual first step weight acceptance ground reaction force (GRF) in GI as we observed in this study and is consistent with the findings of other studies (Ellmers et al., 2020).

Other PCA results from the first step of GI in this study, show that fallers have a more gradual rise in vertical force during weight acceptance. This is consistent with Kwon et al., (2018) who observed that fallers take a longer time to reach maximum weight acceptance and a longer time to mid-stance than non-fallers. They suggested that this could be related to a longer initial stance time associated with increased double limb stance. While we did not observe the longer time to mid-stance in our PCA results, the longer initial stance time could be suggestive of a longer double support stance time as well as a more tentative initial contact. A more tentative initial contact would be consistent with the PCA kinetic results during weight acceptance showing significant patterns of lower medial forces to minimize ML instability and higher anterior forces to decelerate the whole body. This would also support the shorter step length observed in the fallers compared to non-fallers. The PCA results during mid-late stance also showed a significant pattern of lower posterior forces in the fallers compared to the non-fallers which could suggest a reduced push-off force. It has been observed that older adults have reduced push-off force as compared to young adults (Yamada et al., 1988). Perhaps the faller cohort in this study also have further reduced push-off compared to the non-faller cohort which could be related to a more tentative gait or decreased strength to support forward progression. However, such interpretations are limited given the limited research of GI GRF data comparing older adult fallers and non-fallers (Kwon et al., 2018).

Overall, this study demonstrated the combined value of time-series PCA alongside discrete variable analysis to characterize differences in COP kinematics and GRFs between fallers and non-fallers in GI. This analysis corroborated previous findings and forms a more comprehensive comparison of fallers and non-fallers during GI that may improve clinical interpretation and the clinical relevance of this task for fall-risk assessment. Further, the results of the analysis highlight the importance of assessing the COM movement alongside COP kinematics and kinetics to best characterize differences as well as functionally interpret balance control strategy differences between fallers and non-fallers during GI. The use of motion capture and or waist-worn inertial measurement units could add greater insight into the assessment of this task and potentially further improve the clinical relevance and accessibility of this analysis.

Limitations and future directions

All participants were recruited from the Greater Victoria area in British Columbia, Canada and were relatively young (77.8 ± 5.0 years) and in generally good health. Therefore, future studies with a larger, more diverse sample would provide greater generalizability. Future studies may also consider adding waist-worn inertial measurement units to concurrently assess the COM movements alongside COP movements during gait initiation to provide further insights on fall risk differences. Further, assessments of GI under dual-task conditions may show greater differences between fallers and non-fallers, when comparing intra-interindividual difference scores (Commandeur et al., 2018).

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