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Investigating the Relationship Between Falls-Risk, Static Balance and Dynamic Balance in Older Adults.

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Abstract

Background: Falls among older adults are influenced by interacting physical, cognitive, behavioural, health-related, and environmental factors, making their identification and assessment complex. Questionnaire-based screening, static balance assessment, and dynamic gait analysis can identify different indicators of falls-related vulnerability. However, these approaches are often considered separately, and comparatively little research has examined whether questionnaire-based screening and static balance assessment classify falls-risk similarly, or whether poorer quiet-standing control is reflected in dynamic gait asymmetry. This raises the overarching research question: To what extent do these assessment approaches provide overlapping or complementary information about falls-related vulnerability in community-dwelling older adults?

Aims: This study examined the relationships among questionnaire-based falls-risk screening, static balance assessment, and gait asymmetry in community-dwelling older adults. Specifically, it compared STEADI falls-risk classification with force-plate-derived static balance classification, examined whether static balance-derived risk groups differed in gait asymmetry following brief periods of mobility and immobility, and assessed whether selected gait-asymmetry variables contributed to static balance-derived classification.

Methods: Thirty-three community-dwelling older adults aged 73.94 ± 4.83 years participated in a within-subject correlational study. Falls risk was assessed using the STEADI Stay Independent questionnaire, with participants classified as Low- or High-risk. Static balance was assessed using a portable BTrackS force plate during three 20-second eyes-closed quiet-standing trials. Centre-of-pressure outcomes included path length, sway velocity, frequency, spatial dispersion, directional displacement, and regularity. The proprietary static balance classification was consolidated into Low- and High-risk groups for comparison with STEADI.

Dynamic balance was assessed using Loadsol® instrumented insoles during two self-paced 10-metre walking tasks: a Mobile-10MWT following one minute of walking, and an Immobile-10MWT following one minute of sitting. Spatiotemporal and kinetic variables were collected for each limb, and absolute symmetry index values were calculated to quantify gait asymmetry. Agreement between STEADI and static balance classifications was examined using McNemar's test and Cohen's kappa. Group and condition effects on gait asymmetry

were examined using two-way mixed-effects analyses, and binary logistic regression assessed whether selected gait-asymmetry variables predicted static balance-derived falls-risk classification.

Results: STEADI classified 27 participants as Low-risk and six as High-risk, whereas the static balance assessment classified 12 as Low-risk and 21 as High-risk following category consolidation. The classifications differed significantly, $\chi^2(1) = 11.53$, $p = 0.0007$, and demonstrated only slight agreement (k = 0.12). Participants classified as High-risk by the static assessment demonstrated greater centre-of-pressure path length (46.62 ± 8.86 vs 24.92 ± 3.78 cm, p < 0.0001), velocity (2.34 ± 0.44 vs 1.23 ± 0.19 $cm \cdot s^{-1}$, $p < 0.0001$), frequency, spatial dispersion, directional displacement, and anterior-posterior approximate entropy than the Low-risk group.

Gait asymmetry did not differ between the Mobile-10MWT and Immobile-10MWT for any spatiotemporal or kinetic variable, and no test-condition-by-group interactions were identified. However, the static balance-derived High-risk group demonstrated greater asymmetry in ground contact time, swing time, peak impact time, midstance time, and average loading rate across the two conditions. No group differences were identified for the remaining spatiotemporal or kinetic outcomes. Mean swing-time and average loading-rate asymmetry jointly contributed to a significant logistic regression model, $\chi^2(2) = 12.393$, $p = 0.002$. The model had a Nagelkerke R^2 of 0.429 and correctly classified 75.8% of participants, although neither predictor was independently significant.

Conclusion: Questionnaire-based screening, static balance assessment, and dynamic gait analysis provided related but distinct information in this cohort. The STEADI tool identified fewer participants as High-risk compared to static balance assessments, with some Low-risk individuals showing elevated sway. However, without monitoring falls, these cases can't be confirmed as false negatives. Those classified as High-risk by the static assessment displayed greater asymmetry in certain gait aspects, though this wasn’t consistent across all measures. Seated immobility for one minute didn’t affect gait asymmetry in the tested conditions. These findings indicate that combining objective performance measures with questionnaire screenings may improve falls-risk assessments by highlighting postural control and gait asymmetry not captured through self-report. Future research is needed to link these outcomes to actual falls.

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List of Abbreviations

A

ACC = Accident Compensation Corporation

A-P = Anterior-posterior

ASI = Absolute symmetry index

ASI% = Absolute symmetry index percentage

AVLR = Average loading rate

B

BBS = Berg Balance Scale

BMI = Body mass index

BTrackS = Balance Tracking Systems

BW = Body weight

C

CoP = Centre of pressure

F

FRAT = Falls-Risk Assessment Tool

FROP-Com = Falls-Risk for Older People in the Community

H

HQSC = Health Quality & Safety Commission

Hz = Hertz

I

ICC = Intraclass correlation coefficient

Immobile-10MWT = Immobile 10-Metre Walk Test

IMU = Inertial measurement unit

M

M-L = Mediolateral

Mobile-10MWT = Mobile 10-Metre Walk Test

N

NIH = National Institutes of Health

R

RMS = Root-mean-square

S

SD = Standard deviation

SI = Symmetry index

SI% = Symmetry index percentage

SPSS = Statistical Package for the Social Sciences

spm = Strides per minute

STEADI = Stopping Elderly Accidents, Deaths and Injuries

T

TUG = Timed Up and Go

W

WHO = World Health Organisation

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Chapter One: Introduction

Population ageing has increased the importance of maintaining mobility, independence, and quality of life in later adulthood. Although ageing is highly individual, changes in sensory function, muscle strength, cognitive processing, and postural control can reduce the capacity to maintain stability during everyday activities (Beard et al., 2016; Peterka, 2018; Sturnieks et al., 2008). Balance is required during standing, walking, turning, transferring between positions, and responding to unexpected disturbances (Horak, 2006; Shumway-Cook et al., 2000; Sturnieks et al., 2008). When the demands imposed by a task or environment exceed an individual's available balance capacity, vulnerability to instability and falling may increase (Maki & McIlroy, 2006; Montero-Odasso et al., 2022).

Falls are a major public-health concern among older adults because they are associated with injury, disability, reduced confidence, activity restriction, loss of independence, hospitalisation, and mortality (James et al., 2020; Montero-Odasso et al., 2022; World Health Organisation [WHO], 2021b). Their effects also extend beyond the individual, creating substantial social and economic demands for families, communities, and healthcare systems (Florence et al., 2018; WHO, 2021b). In Aotearoa New Zealand, the Accident Compensation Corporation (ACC) reports accepting approximately 155,000 claims each year for falls and fracture-related injuries among people aged 65 years and over, at an annual recovery cost of approximately NZ$195 million (ACC, 2021). This burden highlights the importance of identifying falls-related vulnerability before injury or loss of independence occurs.

Current falls-prevention guidelines recommend a risk-stratification approach with two principal entry pathways: opportunistic case-finding during routine healthcare contact and assessment following presentation with a fall or fall-related injury (Montero-Odasso et al., 2022). During opportunistic case-finding, older adults may first be asked about falls within the previous 12 months, unsteadiness, and concern about falling. A positive response prompts further assessment of gait or balance, which, together with falls history and the severity of any presenting fall, informs classification as having Low-, Intermediate-, or High-risk and the level of subsequent assessment and intervention. Consequently, access to objective gait or balance assessment within this pathway may depend on the information reported during initial case-finding. This is important because the guidelines recognise that

asking only about previous falls has limited sensitivity, does not account for other common falls-risk factors, and may therefore result in false-negative classifications (Montero-Odasso et al., 2022). The need to consider information beyond falls history reflects the broader complexity of falls-risk, as falls rarely arise from a single cause. Falls-risk reflects the interaction of physical capacity, health status, cognition, behaviour, psychological factors, environmental exposure, and task demands (Deandrea et al., 2010; Delbaere et al., 2010; Montero-Odasso et al., 2022). Accordingly, individual screening tools or balance and mobility assessments may not represent the full range of factors contributing to falls-risk (Lusardi et al., 2017; Montero-Odasso et al., 2022). Falls-risk assessment may therefore draw on multiple sources of information, including self-reported risk indicators and performance during standing and movement.

Questionnaire-based screening provides a practical means of identifying recognised falls-risk factors, including previous falls, unsteadiness, concern about falling, medication use, and functional difficulty (Montero-Odasso et al., 2022; Rubenstein et al., 2011). However, the information obtained may be influenced by several factors. Retrospective reporting is imperfect, as some older adults do not recall falls that occurred during specific periods within the preceding three to 12 months, and fall-related injuries may also be forgotten (Cummings et al., 1988), while responses may also depend on how individuals interpret a fall and whether they perceive themselves to be at risk (Yardley et al., 2006; Zecevic et al., 2006). In addition, some older adults may be reluctant to disclose falls-related concerns because of embarrassment or concern about being perceived as frail or dependent (McInnes & Askie, 2004; Yardley et al., 2006). Objective assessment can therefore complement questionnaire-based screening by providing direct information about balance performance during specific tasks.

Balance may be assessed in both relatively stationary and movement-based contexts, reflecting different demands on postural control (Horak, 2006; Pollock et al., 2000; Sturnieks et al., 2008). Force platforms can quantify centre-of-pressure behaviour during quiet standing, providing information about postural sway and the corrective adjustments used to maintain stability (Prieto et al., 1996; Winter, 1995). Portable force-plate systems have also increased the accessibility of these measurements outside specialised laboratory settings (Goble & Baweja, 2018; O'Connor et al., 2016). Static standing, however, represents only one

context of balance control. Many falls occur during movement-based activities, including walking and transferring between positions, and may involve slips, trips, or inappropriate weight shifting (Maki, 1997; Robinovitch et al., 2013).

Dynamic balance requires stability to be maintained while the centre-of-mass and base of support are changing (Winter, 1995). During walking, this depends on the coordinated control of step timing, limb loading, foot placement, and adaptation to changing movement demands (Maki & McIlroy, 2006; Perry & Burnfield, 2010). Instrumented insoles allow these features to be examined across consecutive steps by quantifying temporal and vertical loading characteristics for each limb (Razak et al., 2012; Renner et al., 2019; Seiberl et al., 2018). This provides an opportunity to identify between-limb differences in walking performance that may not be apparent from broader clinical measures such as overall walking time or speed (Nakanowatari, Hoshi, Sone, et al., 2024).

Falls-risk screening, static balance assessment, and dynamic gait analysis may provide related but non-equivalent information. Research directly comparing static and dynamic balance indicates that associations between them are often small or non-significant, suggesting that these assessments should not be assumed to be interchangeable (Kiss et al., 2018; Rizzato et al., 2021). However, it remains unclear how questionnaire-based falls-risk classifications correspond with objective static balance-derived classifications, and how information obtained from standing and movement-based assessments relates within independently mobile older adults. This raises the overarching research question: To what extent do these assessment approaches provide overlapping or complementary information about falls-related vulnerability in community-dwelling older adults?

This thesis will present a review of the literature relating to falls-risk, balance control, and the assessment of static and dynamic balance in older adults. Chapter Two will examine falls-risk in older adults, the sensory, cognitive, and neuromuscular systems that support balance, and the distinction between static and dynamic balance. Falls-risk screening, clinical balance tests, force-platform assessment, and instrumented gait analysis will then be reviewed, leading to consideration of the relationships among falls-risk screening, static balance, and dynamic balance measures. Based on this literature, the research aims and hypotheses will be presented in Chapter Three. Chapter Four will describe the methodology, followed by the results in Chapter Five. Chapter Six will discuss the findings, limitations, and directions for

future research. Finally, Chapter Seven will present the overall conclusions and practical implications of the thesis.

Chapter Two: Literature Review

2.1 Falls-Risk in Older Adults

Falls-risk provides the applied context for the present thesis, as falls are associated with injury, disability, reduced independence, and substantial healthcare burden among older adults (James et al., 2020; Montero-Odasso et al., 2022; WHO, 2021b). Before examining balance mechanisms and assessment methods, it is important to define the older adult population, clarify the concepts of falls and falls-risk, and outline the multifactorial contributors to falling. This section, therefore, introduces the older adult population, describes the clinical relevance and burden of falls, summarises key falls-risk factors, and examines fear of falling and falls self-efficacy.

2.1.1 Defining Older Adults

Understanding falls-risk, balance, and gait in later life first requires clarity about how "older adults" are defined in the research literature. Research examining these domains commonly classifies older adults as individuals aged 65 years and over, reflecting population-level trends in age-related physiological change rather than a strict biological threshold (Beard et al., 2016; National Institutes of Health [NIH], 2000; Orimo et al., 2006; Sturnieks et al., 2008). This convention also aligns with international ageing classifications and the age thresholds used by many health and support services worldwide, including superannuation entitlements, residential aged-care funding, community nursing services, and fall-prevention programmes (Beard et al., 2016; Cesari et al., 2017; Fuller, 2000; NIH, ; Rula et al., 2011). In research practice, the 65 years and over threshold therefore provides a consistent and comparable framework across epidemiological, clinical, and community settings. However, the World Health Organisation (WHO) uses both 60 years and over and 65 years and over, depending on context (WHO, 2015). The 60 years and over classification is typically applied in global reporting to allow comparability across countries with differing life expectancies and demographic structures, whereas 65 years and over is commonly used in clinical, epidemiological, and high-income health-system contexts, including fall prevention, functional decline, and age-related epidemiology (Beard et al., 2016; Cesari et al., 2017; Sturnieks et al., 2008; WHO, 2015). Consistent with research conventions in New Zealand, Australia, Europe, and the United States, and with the majority of studies examining postural

control and falls-risk, the present thesis adopts the 65 years and over definition (Beard et al., 2016; Cesari et al., 2017; Fuller, 2000; Sturnieks et al., 2008).

Contemporary gerontology and public-health organisations also emphasise the importance of respectful, person-centred terminology when referring to individuals aged 65 years and over (Beard et al., 2016; NIH, ; WHO, 2021a). The term “elderly” is increasingly avoided in research and practice because it can imply frailty, dependency, or homogeneity, and may contribute to ageist stereotypes (Beard et al., 2016; Calder, 2022; NIH, ; WHO, 2021a). In contrast, terms such as “older adults” or “older people” are preferred as they better reflect the diversity, capability, and agency of individuals in later life. For this reason, the present thesis will use the term “older adults” when referring to individuals aged 65 years and over. This terminology supports a more accurate and dignifying representation of ageing populations and aligns with international guidance and national organisations (Beard et al., 2016; Calder, 2022; NIH, ; WHO, 2021a).

However, chronological age alone does not fully represent functional ability. Studies show substantial variation in sensory capacity, muscle strength, gait stability, and falls-risk among people of the same age (Beard et al., 2016; Bellumori et al., 2013; Charlier et al., 2016; Larsson et al., 2016; Peterka, 2018). Many adults in their late sixties remain highly active and stable, while others experience earlier declines due to multimorbidity, vestibular impairment, neurological conditions, or reduced physical activity (Cesari et al., 2017; Kristinsdottir et al., 2001; Maki et al., 1994; Manckoundia et al., 2008; Stelmach et al., 1989). Research in community-dwelling older adults consistently demonstrates that balance performance varies widely even within the same age group due to differences in sensory integration, postural adaptation, and neuromuscular capacity (Lord & Ward, 1994; Manchester et al., 1989; Teasdale et al., 1991).

For these reasons, research increasingly emphasises physiological and functional ageing alongside chronological age. Physiological ageing refers to age-related changes in body systems that contribute to an individual’s intrinsic capacity, including sensory, locomotor, cognitive, psychological, and vitality-related domains (Cesari et al., 2017; WHO, 2015; WHO, 2020). Functional ageing refers to how these capacities interact with the environment to influence an individual’s ability to perform daily tasks, maintain mobility, and respond to balance challenges (WHO, 2015; WHO, 2020). This distinction is especially important in

balance and falls research, where individuals of the same chronological age can differ markedly in postural stability and fall-risk. These differences are clinically important because they shape how falls and falls-risk are defined, identified, and understood in older adults.

2.1.2 Definition and Clinical Relevance of Falls and Fall-Risk

Following the definition of the older adult population, it is important to clarify how falls are defined, as this influences how falls are reported, recorded, and interpreted in both research and clinical practice. The WHO defines a fall as: “inadvertently coming to rest on the ground, floor or other lower level, excluding intentional change in position to rest in furniture, wall or other objects” (WHO, 2008, p. 1). This aligns with the Prevention of Falls Network Europe consensus definition, which recommends a fall to be described as: “an unexpected event in which the participants come to rest on the ground, floor, or lower level.’” (Lamb et al., 2005, p. 1619). Thus, these definitions emphasise that falls are unintended events and distinguish them from deliberate changes in position, such as sitting or lying down voluntarily (Lamb et al., 2005; WHO, 2008).

Standardised definitions are important because older adults, clinicians, and researchers may interpret the term “fall” differently. Older adults may describe a fall in relation to loss of balance, embarrassment, injury, or the circumstances surrounding the event, whereas clinical and research definitions often focus on the final outcome of coming to rest on the ground or a lower level (Lamb et al., 2005; WHO, 2008; Zecevic et al., 2006). Zecevic et al. (2006) showed that perceptions of falls can differ between older adults, healthcare providers, and the research literature, highlighting the importance of using clear operational definitions when asking older adults about previous falls. Without a standard definition, falls may be under-reported, misclassified, or inconsistently recorded across studies and clinical settings (Lamb et al., 2005; WHO, 2008; Zecevic et al., 2006).

It is also important to distinguish between a fall and fall-risk. A fall is an observed event that has already occurred, whereas fall-risk refers to the likelihood that an individual will experience a fall in the future (Lamb et al., 2005; Montero-Odasso et al., 2022). The World Falls Guidelines describe falls-risk stratification as a process for grading an individual’s risk of falling and guiding further assessment or intervention (Montero-Odasso et al., 2022). This distinction is important because falls-risk assessment is predictive rather than purely

descriptive. It does not simply document whether a fall has occurred; rather, it considers whether an older adult has factors that increase the likelihood of future falls (Lusardi et al., 2017; Montero-Odasso et al., 2022; Phelan et al., 2015).

Fall-risk is closely related to balance, but it is not the same as balance impairment. Balance deficits, gait instability, reduced strength, and impaired sensory function may increase fall-risk, but falls are multifactorial events influenced by the interaction of intrinsic, extrinsic, behavioural, and situational factors (Deandrea et al., 2010; Montero-Odasso et al., 2022; WHO, 2008). For example, an older adult with reduced balance capacity may not fall in a familiar, well-lit environment, but may become vulnerable when walking on uneven ground, rushing, carrying objects, or responding to a sudden perturbation (Maki, 1997; Maki & McIlroy, 2006; Sturnieks et al., 2008). Conversely, an individual may have relatively good balance performance but still be at increased risk because of medication effects, visual impairment, cognitive impairment, environmental hazards, or previous falls (Deandrea et al., 2010; Lusardi et al., 2017; Montero-Odasso et al., 2022).

The clinical relevance of falls is substantial because a fall can lead to injury, loss of confidence, reduced mobility, hospitalisation, institutionalisation, and mortality (Lamb et al., 2005; Montero-Odasso et al., 2022; WHO, 2008). Montero-Odasso et al. (2022) distinguish among falls, recurrent falls, severe falls, and fall-related injuries, reflecting that not all falls have the same clinical consequences. Even when a fall does not result in serious physical injury, it may still contribute to fear of falling, reduced activity, loss of independence, and reduced quality of life (Lamb et al., 2005; Montero-Odasso et al., 2022; WHO, 2008). Therefore, falls are clinically important not only because of the immediate risk of injury, but also because of their potential to initiate a longer trajectory of functional decline.

In older adults, the importance of fall-risk identification lies in prevention. The World Falls Guidelines recommend that clinicians routinely ask older adults about falls, because falls are often not spontaneously reported, and that those reporting falls, gait or balance disturbance, dizziness, or concerns about falling should receive further assessment where appropriate (Montero-Odasso et al., 2022). This approach reflects the need to identify risk before falls lead to injury or loss of independence. While defining falls and fall-risk establishes the clinical problem, the broader impact of falls extends beyond the fall event

itself. The following section, therefore, examines the physical, psychological, social, and economic burden of falls in older adults.

2.1.3 Burden of Falls in Older Adults

Falls in older adults are a major public-health concern because they contribute substantially to morbidity, mortality, disability, healthcare use, and loss of independence (James et al., 2020; Montero-Odasso et al., 2022; WHO, 2021b). Globally, the WHO reports that more than 684,000 people die each year as a result of falls, while approximately 172 million people experience short- or long-term disability following a fall (WHO, 2021b). Falls are also the world's second leading cause of injury-related mortality and account for more than 38 million disability-adjusted life years each year (WHO, 2021b). These estimates are consistent with the Global Burden of Disease Study 2017, which estimated that falls accounted for 695,771 deaths and approximately 35.9 million disability-adjusted life years globally in 2017 (James et al., 2020).

The impact of falls increases markedly with age and is reflected in healthcare use across older populations. In Aotearoa New Zealand, the Health Quality and Safety Commission (HQSC) reported that, in 2016, approximately 216,000 people aged 50 years and over had an accepted ACC claim for a fall-related injury, and 27,090 people aged 50 years and over attended hospital following a fall (HQSC, 2018). Hospital-attendance rates increased from 12.9 per 1,000 people aged 65-74 years, to 36.4 per 1,000 people aged 75-84 years, and 102.6 per 1,000 people aged 85 years and over (HQSC, 2018). Similar patterns are reported internationally. In the United States, falls among older adults result in approximately three million emergency department visits and one million fall-related hospitalisations each year (CDC, 2026). Hip fractures represent a particularly important fall-related injury, with nearly 319,000 older adults hospitalised for hip fractures each year in the United States, and 88% of emergency department visits and hospitalisations for hip fractures in 2019 were attributed to falls (Moreland et al., 2023). In Western Europe, Haagsma et al. (2020) reported that 11.7 million adults aged 70 years and over sought medical attention for injury in 2017, of which 8.4 million, or 71.9%, were due to fall-related injury. These findings demonstrate that falls represent a major source of injury and healthcare demand across ageing populations.

The physical consequences of falls range from bruising and soft-tissue injury to fractures, head injuries, hospitalisation, and death (Ganz et al., 2005; James et al., 2020; Montero-Odasso et al., 2022; WHO, 2021b). Although many falls do not result in severe injury, serious injuries are clinically important because they can initiate a decline in mobility, independence, and functional capacity. Ganz et al. (2005) reported that 5-10% of falls result in serious injuries such as head injury, serious laceration, or fracture. Hip fractures and head injuries are particularly important because they are associated with acute medical care, rehabilitation needs, reduced mobility, and increased risk of long-term support or institutional care (Montero-Odasso et al., 2022; WHO, 2021b). Therefore, the physical consequences of falls include not only the immediate injury, but also the functional decline that may follow.

Falls also have psychological and social consequences that may occur even when no major physical injury is sustained. Lamb et al. (2005) identified psychological outcomes, physical activity, and health-related quality of life as important categories in fall-injury prevention trials, demonstrating that fall outcomes extend beyond injury counts alone. A fall may reduce confidence, increase concern about falling, and contribute to activity avoidance (Jørstad et al., 2005; Lamb et al., 2005; Zijlstra et al., 2007). Zijlstra et al. (2007) reported that fear of falling and activity avoidance are common among community-living older adults and are associated with functional decline, restricted social participation, reduced quality of life, increased risk of falling, and institutionalisation. Reduced participation may also increase reliance on family, caregivers, or health services for daily activities and mobility support (WHO, 2021b). In this way, the consequences of falls can extend beyond the individual affecting wider social networks and care systems.

The economic burden of falls reflects the costs of emergency care, hospital admission, rehabilitation, long-term care, compensation, and ongoing support needs (Florence et al., 2018; Scuffham et al., 2003; WHO, 2021b). In the United States, Florence et al. (2018) estimated that medical costs attributable to fatal and non-fatal falls in older adults were approximately US$50 billion in 2015, with most costs associated with non-fatal falls. In the United Kingdom, Scuffham et al. (2003) estimated that falls in people aged 60 years and over cost the government £981 million, with inpatient admissions and long-term care representing major cost drivers. In Aotearoa New Zealand, ACC reports that approximately

155,000 claims are accepted each year for falls and fracture-related injuries among people aged 65 years and over, costing approximately NZ$195 million to support recovery (ACC 2021). More recent ACC data indicate that fall-related injuries continue to generate a substantial number of new and active claims each year, with the cost of active fall-related claims exceeding NZ$2.5 billion in 2025 (ACC, 2026). These figures are relevant to the present thesis because the study focuses on falls-risk assessment in community-dwelling older adults in Aotearoa New Zealand.

Overall, falls impose physical, psychological, social, and economic consequences for older adults, families, communities, and health systems. This burden highlights the importance of identifying older adults at increased risk before injury, activity restriction, or loss of independence occurs. Effective prevention, however, requires an understanding of the interacting factors that increase an individual's vulnerability to falling.

2.1.4 Risk Factors for Falls

Falls in older adults are multifactorial and usually arise from the interaction of several risk factors rather than from one isolated cause. These risk factors are commonly grouped into intrinsic, extrinsic, behavioural, psychological, and situational categories, which together influence an individual's vulnerability to falling and the likelihood that they will encounter situations that challenge stability (Deandrea et al., 2010; Montero-Odasso et al., 2022; WHO, 2021b). Falls-risk, therefore, reflects the interaction between physical capacity, health status, environmental exposure, behaviour, confidence, and task demands (Delbaere et al., 2010; Montero-Odasso et al., 2022; WHO, 2021b).

Intrinsic risk factors are characteristics within the individual that increase vulnerability to falling. These include increasing age, previous falls, impaired gait and balance, reduced muscle strength, sensory impairment, cognitive impairment, chronic disease, frailty, pain, medication use, and urinary symptoms (Deandrea et al., 2010; Delbaere et al., 2010; Montero-Odasso et al., 2022; Muir et al., 2012; Stubbs et al., 2014). Among these factors, previous falls are particularly important because they provide evidence that the individual has already experienced difficulty maintaining stability in daily life (Deandrea et al., 2010). In a systematic review and meta-analysis of prospective studies in community-dwelling older adults, Deandrea et al. (2010) reported that individuals with a history of falls had

substantially higher odds of future falls, with the association stronger for recurrent falls than for single falls. A previous fall should therefore be interpreted not only as a past event, but also as an important marker of potential ongoing vulnerability.

Mobility-related impairments are also central to intrinsic falls-risk. Deandrea et al. (2010) reported that older adults with gait problems had approximately twice the odds of falling compared with those without gait problems, both for any falls and for recurrent falls. The World Falls Guidelines similarly include gait and balance, muscle strength, walking-aid use, foot problems, and footwear among the factors that should be considered within multifactorial falls-risk assessment (Montero-Odasso et al., 2022). These factors are clinically relevant because they influence an individual's ability to respond to perturbations, negotiate uneven surfaces, transition between positions, and maintain stable walking during daily activities (Maki, 1997; Maki & McIlroy, 2006; Sturnieks et al., 2008). In this context, walking-aid use is best interpreted as part of a wider mobility profile, as it may indicate underlying weakness, impaired balance, reduced confidence, or functional limitation (Deandrea et al., 2010; Montero-Odasso et al., 2022).

Cognitive impairment, medication use, and broader health status further contribute to intrinsic vulnerability. Safe mobility requires attention, planning, judgement and the ability to adapt movement to changing environmental demands (Mirelman et al., 2012; Manuel Montero-Odasso et al., 2012; Muir et al., 2012). Muir et al. (2012) reported that both global cognitive impairment and executive-function impairment were associated with increased falls-risk, highlighting the role of attention, inhibition, task switching, and movement planning during everyday mobility. Medication use and polypharmacy are also recognised contributors to falls-risk, particularly when medications affect alertness, blood pressure, reaction time, dizziness, or postural control (Blain et al., 2018; Deandrea et al., 2010; Montero-Odasso et al., 2022). Other health-related contributors, including vertigo, Parkinson's disease, frailty, urinary incontinence, sensory impairment, and pain, may further reduce the capacity to move safely and respond effectively to instability (Deandrea et al., 2010; Montero-Odasso et al., 2022; Stubbs et al., 2014).

Extrinsic risk factors refer to hazards in the physical environment that increase the likelihood of a fall. These include slippery or uneven surfaces, poor lighting, clutter, loose rugs, stairs without appropriate rails, unsafe bathrooms, unsuitable footwear, and poorly fitted or

inappropriate assistive devices (Clemson et al., 2023; Montero-Odasso et al., 2022; Stark et al., 2021; WHO, 2021b). Environmental hazards are clinically important because they can increase the demands placed on an individual's available physical capacity, particularly when intrinsic risk factors such as reduced strength, sensory impairment, slower reaction time, or impaired gait are also present (Clemson et al., 2023; Montero-Odasso et al., 2022; Sturnieks et al., 2008). For example, a surface change, poor lighting, or trip hazard may be managed successfully by an older adult with adequate strength, vision, and reactive stepping capacity, but may contribute to a fall when sensory or motor responses are reduced (Maki & McIlroy, 2006; Sturnieks et al., 2008). Importantly, environmental hazards are modifiable. Stark et al. (2021) found that a brief home hazard-removal programme reduced the rate of falls, although the primary time-to-first-fall outcome did not differ significantly between groups. Clemson et al. (2023) also reported that home fall-hazard interventions reduced the overall rate of falls by 26%, with stronger effects among people selected as having a higher falls-risk. These findings suggest that environmental hazards are not only associated with falls but can also be practical targets for prevention when identified and addressed appropriately.

Behavioural and situational risk factors describe how individuals interact with their environment and the specific circumstances in which falls occur. Behavioural factors may include physical inactivity, activity avoidance, rushing, risk-taking, not using prescribed walking aids, wearing inappropriate footwear, or failing to adapt behaviour when conditions are unsafe (Delbaere et al., 2010; Montero-Odasso et al., 2022; WHO, 2021b). These factors are complex because falls-risk is influenced not only by physical capacity, but also by how much challenge or exposure an individual experiences. Delbaere et al. (2010) showed that fall-risk increased when multiple factors occurred together, including poorer physiological function, disability, executive-function impairment, reduced coordinated stability, and activity-related exposure. In their model, estimated fall-risk ranged from 11% in older adults with no identified risk factors to 54% in the highest-risk subgroup, demonstrating that risk increases when physical, cognitive, functional, and activity-related factors combine.

Situational factors refer to the specific tasks or contexts in which balance is challenged. These may include turning, rising from a chair, walking while carrying objects, negotiating stairs or kerbs, walking on uneven ground, moving in low light, dual-task walking, or responding to slips and trips (Maki, 1997; Maki & McIlroy, 2006; Manuel Montero-Odasso et

al., 2012; Montero-Odasso et al., 2022). These situations increase the demands placed on attention, sensory integration, lower-limb strength, reaction time, and gait adaptability, which helps explain why an older adult may appear stable in a controlled setting but remain vulnerable during everyday movement (Maki & McIlroy, 2006; Mirelman et al., 2012; M. Montero-Odasso et al., 2012; Sturnieks et al., 2008).

Psychological factors also contribute to falls-risk, particularly fear of falling, concern about falling, and falls self-efficacy. Fear of falling is commonly described as concern, anxiety, or worry about falling during daily activities, whereas falls self-efficacy refers to an individual's confidence in their ability to perform activities without falling (Hadjistavropoulos et al., 2011; Tinetti et al., 1990; Yardley et al., 2006). These constructs are related but not identical, as an older adult may experience concern about falling, reduced confidence, or both. This distinction is clinically important because psychological responses to falling can influence movement behaviour, activity choices, and participation (Hadjistavropoulos et al., 2011; Tinetti et al., 1990; Yardley et al., 2006).

Fear of falling may occur after a fall, but it is not limited to individuals with a previous fall history (Hadjistavropoulos et al., 2011). It may reflect perceived vulnerability, declining physical capacity, reduced balance confidence, or awareness of environmental and task-related challenges. Older adults who are concerned about falling may restrict walking, avoid community participation, reduce exercise, or limit activities perceived as risky (Delbaere et al., 2004; Hadjistavropoulos et al., 2011; Zijlstra et al., 2007) Over time, this may contribute to a cycle in which concern leads to activity restriction, activity restriction contributes to deconditioning, and declining function further increases falls-risk (Delbaere et al., 2004; Zijlstra et al., 2007).

Fear of falling may also influence how older adults move. Concern about falling has been associated with more cautious movement strategies, including slower walking, shorter steps, reduced cadence, increased double-support time, and altered postural control (Delbaere et al., 2009; Hadjistavropoulos et al., 2011; Maki, 1997). These adaptations may be protective in some situations, but they may also reduce gait adaptability when a rapid response is required, such as during a trip, slip, obstacle, or change in surface condition (Hadjistavropoulos et al., 2011; Maki, 1997). This demonstrates that psychological factors

interact with physical balance and gait performance, rather than existing separately from them.

Recent evidence further supports the inclusion of concern about falling in falls-risk assessment. Ellmers et al. (2025) reported that baseline concern about falling was an independent predictor of future falls in older adults, and this relationship was observed across different measures of concern, including the Falls Efficacy Scale-International and single-item concern measures. In contrast, balance confidence measured using the Activities-Specific Balance Confidence Scale did not predict future falls in the same way (Ellmers et al., 2025). This contrast suggests that concern about falling and balance confidence should not be treated as interchangeable constructs. Concern about falling may reflect perceived threat, anxiety, avoidance, or behavioural adaptation, whereas balance confidence reflects perceived capability during specific activities (Hadjistavropoulos et al., 2011; Tinetti et al., 1990; Yardley et al., 2005). The finding, therefore, supports the inclusion of concern about falling within multifactorial falls-risk assessment, while also reinforcing the need to interpret psychological measures carefully.

Overall, falls-risk in older adults reflects the interaction between individual capacity, environmental exposure, behavioural choices, psychological responses, and task-specific demands. Intrinsic factors may reduce the capacity available for maintaining stability, extrinsic factors may increase the challenge imposed by the environment, behavioural factors may influence exposure or avoidance, psychological factors may shape confidence and movement behaviour, and situational factors may determine when stability is most strongly challenged (Deandrea et al., 2010; Delbaere et al., 2010; Montero-Odasso et al., 2022; WHO, 2021a). Therefore, these interacting factors support a multidimensional approach to fall risk assessment, rather than relying on a single source of information. Although falls-risk is shaped by many factors, balance remains one of the key mechanisms through which older adults maintain stability and avoid falls. The following section examines balance in older adults, including the sensory, cognitive, neuromuscular, and musculoskeletal systems that support postural control.

2.2 Balance in Older Adults

Balance is central to how older adults maintain independence and safely complete everyday tasks such as standing, walking, turning, transferring, and responding to unexpected disturbances (Horak, 2006; Shumway-Cook et al., 2000; Sturnieks et al., 2008). It reflects the coordinated function of sensory, cognitive, and neuromuscular systems, which together allow the body to detect movement, interpret environmental demands, and generate appropriate postural responses (Peterka, 2018; Sturnieks et al., 2008; Woollacott et al., 1986). With advancing age, changes within these systems may reduce the reserve available to maintain stability, particularly when task or environmental demands increase (Assländer & Peterka, 2014; Peterka, 2018; Sturnieks et al., 2008). Understanding how these age-related changes influence falls-risk first requires a clear definition of balance and the processes involved in maintaining postural control.

2.2.1 Understanding Balance in Older Adults

Balance refers to the ability to regulate the body's position in space so that the centre-of-mass remains controlled relative to the base of support, whether the body is stationary or moving (Bohm et al., 2015; Pollock et al., 2000; Winter, 1995). This is achieved through continuous adjustments in posture that maintain stability during standing, walking, and transitions such as turning or rising from a chair (Manckoundia et al., 2008; Peterka, 2018; Sturnieks et al., 2008).

In standing, balance is maintained through small, automatic movements around the ankles, hips, and trunk that keep the centre-of-mass within the limits of stability. These adjustments are reflected mechanically in the movement of the centre-of-pressure, which represents the point at which ground-reaction forces act beneath the feet (Johansson & Magnusson, 1991; Lázaro et al., 2011; Pollock et al., 2000; Prieto et al., 1996; Winter, 1995). Centre-of-pressure behaviour therefore provides an external mechanical representation of how the body controls sway and distributes load during quiet stance and weight-shifting tasks (Maki et al., 1994; Thapa et al., 1994; Winter, 1995).

During movement, balance also involves coordinating the position of the centre-of-mass with step placement, leg trajectory, and momentum. In walking, this is achieved by actively shifting the centre-of-mass forward and laterally while controlling centre-of-pressure

trajectories under each foot to maintain progression and stability (Hausdorff, 2005; Hausdorff et al., 2001; Maki & McIlroy, 2006; Pollock et al., 2000; Sturnieks et al., 2008). Everyday tasks such as turning, navigating uneven surfaces, or responding to disturbances introduce additional demands on gait stability and require coordinated adjustments in leg support, foot placement, and body orientation (Cromwell & Newton, 2004; Kristinsdottir et al., 2001; Larsson et al., 2016; Perry & Burnfield, 2010). Research consistently shows that maintaining balance during movement places greater demands on sensory integration, neuromuscular control, and anticipatory adjustments than quiet standing (Fransson et al., 1998; Peterka, 2018; Roos & Dingwell, 2013).

Therefore, these processes suggest that balance can be interpreted within two main contexts:

1. Static balance, which involves maintaining stability when the body is relatively still, and
2. Dynamic balance, which involves controlling the centre-of-mass during movement.

These distinctions provide a practical framework for describing how balance is expressed in everyday situations and are widely used in research and clinical assessment of postural control in older populations (Horak, 2006; Mancini & Horak, 2010; Pollock et al., 2000; Sturnieks et al., 2008). However, before static and dynamic balance are discussed in detail, it is necessary to consider the physiological systems that maintain balance. Balance performance depends on the integration of sensory information, central processing, and motor output, all of which may change with ageing.

2.2.2 Determinants of Balance

Balance performance in older adults is influenced by the combined function of sensory, cognitive, and neuromuscular systems (Horak, 2006; Peterka, 2018; Sturnieks et al., 2008). These systems allow the body to detect movement, interpret environmental demands, and generate timely corrective responses during standing, walking, transitions, and unexpected perturbations (Peterka, 2018; Sturnieks et al., 2008; Woollacott et al., 1986). With advancing age, reductions in sensory acuity, slower neural processing, reduced muscle strength, and increased cognitive demand can reduce the efficiency of these responses (Bohm et al., 2015; Hughes et al., 1996; Kristinsdottir et al., 2001; Teasdale et al., 1991; Woollacott et al., 1986).

Functionally, these changes may present as increased postural sway, delayed balance corrections, poorer adaptation to changing environments, cautious gait strategies, and reduced ability to recover from slips, trips, or unexpected disturbances. These changes reduce the reserve available for maintaining balance under challenging conditions and increase susceptibility to falls (Baloh et al., 1993; Biasin et al., 2023; Manckoundia et al., 2008; Sturnieks et al., 2008).

2.2.2.1 Sensory Contributions to Balance

Sensory information provides the foundation for balance control by allowing the central nervous system to detect body position, movement, and orientation relative to the environment. The visual, vestibular, and somatosensory systems each contribute different information, and age-related decline in any of these systems can reduce the accuracy of postural responses (Peterka, 2018; Sturnieks et al., 2008).

Somatosensory and proprioceptive input provide information about body position, joint movement, muscle tension, and contact with the support surface through receptors located in muscles, tendons, joints, and skin (Assländer & Peterka, 2014; Shaffer & Harrison, 2007; Skinner et al., 1984; Winter, 1995). The feet, ankles, and lower legs are particularly important because they provide information about body position relative to the support surface. With ageing, reductions in tactile acuity, vibration sensitivity, and joint-position sense are commonly observed, particularly in the distal lower limbs (Hughes et al., 1996; Kristinsdottir et al., 2001; Lord et al., 1991; Skinner et al., 1984; Sturnieks et al., 2008). Sturnieks et al. (2008) reported that older adults demonstrate dorsiflexion and plantarflexion movement-detection thresholds three to four times larger than younger adults under weight-bearing conditions, indicating reduced proprioceptive sensitivity. These changes reduce the precision of afferent feedback and may impair the ability to detect body sway, surface changes, or foot-placement errors (Hughes et al., 1996; Shaffer & Harrison, 2007; Sturnieks et al., 2008). Functionally, this may contribute to increased postural sway, cautious gait patterns, delayed balance recovery, and greater difficulty adapting to uneven or unstable surfaces (Hughes et al., 1996; Laughton et al., 2003; Woollacott et al., 1986).

Vestibular input provides information about head motion, linear acceleration, and gravitational orientation, supporting gaze stabilisation, head orientation, and postural

equilibrium (Baloh et al., 1993; Peterka, 2018; Sturnieks et al., 2008). Vestibular information becomes especially important when vision is reduced, when the support surface is unstable, or when the head and body are moving during walking. With advancing age, vestibular function declines, reducing the accuracy with which head movement and spatial orientation are detected (Agrawal et al., 2009; Baloh et al., 1993; Ciorba, 2015; Sturnieks et al., 2008). Agrawal et al. (2009) reported that 35.4% of United States adults aged 40 years and older had vestibular dysfunction, and that symptomatic vestibular dysfunction was associated with a 12-fold increase in the odds of falling. These changes may contribute to dizziness, imbalance, spatial disorientation, increased sway, and poorer stability during stance and gait (Agrawal et al., 2009; Baloh et al., 1993; Ciorba, 2015). Vestibular asymmetry has also been linked with impaired gait stability and increased falls-risk in older adults. Hansson and Magnusson (2013) found that 13 of 21 older adults who fell had vestibular asymmetry, and all six participants who experienced three or more falls showed vestibular asymmetry. Larsson et al. (2016) also reported that vestibular asymmetry was associated with increased double-support time variability during gait. Therefore, these findings indicate that vestibular impairment may contribute to unstable gait and recurrent falls (Agrawal et al., 2009; Hansson & Magnusson, 2013; Larsson et al., 2016).

Vision provides external reference frames that support spatial orientation, motion perception, obstacle detection, and anticipatory postural control (Baloh et al., 1993; Owsley, 2011; Peterka, 2018). Visual cues inform the central nervous system about environmental layout, surface boundaries, moving objects, and potential hazards, allowing posture and movement to be adjusted before instability occurs (Assländer & Peterka, 2014; Owsley, 2011). With ageing, declines in visual acuity, contrast sensitivity, peripheral vision, and depth perception can impair the precision and speed of visual processing (Chetty et al., 2013; Owsley, 2011). These changes can make it harder to detect obstacles, judge surface changes, or orient the body accurately in space, particularly in low-light or visually complex environments (Owsley, 2011). Sturnieks et al. (2008) reported that postural sway area increases by approximately 30% when visual input is removed by closing the eyes, highlighting the role of vision in maintaining upright stance. For older adults, vision can help compensate for reduced somatosensory or vestibular reliability; however, this reliance also increases vulnerability when visual information is degraded or misleading (Peterka, 2018;

Sturnieks et al., 2008; Teasdale et al., 1991). Therefore, sensory decline can reduce the quality of information available for balance control. The next section examines how the central nervous system integrates and reweights these sensory inputs when task or environmental demands change.

2.2.2.2 Sensory Integration and Reweighting

Effective postural control requires the central nervous system to integrate visual, vestibular, and somatosensory information and dynamically adjust the relative contribution of each input according to the task and environment (Assländer & Peterka, 2014; Peterka, 2018; Pollock et al., 2000; Winter, 1995). Although somatosensory input from the feet and ankles is often dominant during quiet standing on a firm surface, the relative weighting of each sensory system changes when conditions become more challenging (Assländer & Peterka, 2014; Peterka, 2018; Sturnieks et al., 2008). For example, when standing on an unstable surface or moving in low light, the central nervous system must reduce reliance on unreliable information and increase reliance on the remaining sensory inputs to preserve equilibrium (Assländer & Peterka, 2014; Fransson et al., 1998; Peterka, 2018).

Ageing can reduce the efficiency of this adaptive process. Older adults often show slower or less complete sensory reweighting, meaning they may have greater difficulty adjusting when sensory information becomes inaccurate, reduced, or conflicting (Assländer & Peterka, 2014; Kristinsdottir et al., 2001; Teasdale et al., 1991). Teasdale et al. (1991) demonstrated that when visual input changed suddenly, older participants displayed larger sway amplitudes and slower compensatory responses than younger adults, indicating reduced adaptive sensory integration. Assländer and Peterka (2014) demonstrated that sensory perturbation changes produced substantial reweighting effects, with gain changes ranging from a 2.2-fold increase to a 3.6-fold decrease, while high-to-low perturbation transitions showed approximately five times longer time constants than low-to-high transitions. These findings show that balance control depends not only on the availability of sensory information, but also on the speed and effectiveness with which sensory weighting is adjusted.

Visual, vibratory, and vestibular perturbation studies further indicate that older adults may demonstrate delayed or reduced corrective responses when sensory information is disrupted (Fransson et al., 2001; Johansson et al., 2001; Kristinsdottir et al., 2001).

Collectively, these findings suggest that postural instability in older adults often reflects reduced sensory integration and adaptation, rather than a failure of a single isolated sensory system (Assländer & Peterka, 2014; Peterka, 2018; Sturnieks et al., 2008). This is relevant to falls-risk because everyday environments frequently require rapid sensory reweighting, such as walking in dim lighting, turning the head while moving, or stepping onto uneven ground. When this adaptive capacity is reduced, even small environmental challenges may be more likely to destabilise balance (Maki & McIlroy, 2006; Sturnieks et al., 2008). However, the ability to maintain balance in these situations also depends on how sensory information is processed, prioritised, and integrated with motor responses (Horak, 2006; Peterka, 2018; Woollacott & Shumway-Cook, 2002). Central processing and cognitive regulation, therefore, play an important role in postural control, particularly when older adults must divide attention, adapt to changing environments, or respond to complex movement demands.

2.2.2.3 Central Processing and Cognitive Regulation

The central nervous system integrates sensory information and transforms it into coordinated motor output that supports postural orientation, movement planning, and balance recovery (Horak, 2006; Peterka, 2018; Woollacott & Shumway-Cook, 2002). Functionally, balance requires rapid interpretation of sensory information, selection of an appropriate motor response, and allocation of attention when conditions become complex. These processes are especially important during dynamic tasks such as turning, obstacle negotiation, dual-task walking, or recovering from perturbations (Herman et al., 2010; Yogev-Seligmann et al., 2008).

With ageing, central processing speed, executive function, and attentional flexibility may decline, reducing the ability to adapt balance responses quickly (Seidler et al., 2010; Woollacott & Shumway-Cook, 2002; Yogev-Seligmann et al., 2008). Behavioural and neuroimaging research suggests that older adults often recruit greater cognitive resources during postural tasks, reflecting a shift from more automatic balance control toward more attention-demanding regulation (Bohm et al., 2015; Seidler et al., 2010; Woollacott & Shumway-Cook, 2002). This compensation may help maintain stability under simple conditions, but leaves fewer attentional resources available when balance must be maintained alongside another task (Seidler et al., 2010; Wollesen & Voelcker-Rehage, 2014; Woollacott & Shumway-Cook, 2002)

When attentional resources are divided, such as during dual-task walking or cognitive interference, postural and gait performance can deteriorate (Herman et al., 2010; Wollesen & Voelcker-Rehage, 2014; Woollacott & Shumway-Cook, 2002; Yogev-Seligmann et al., 2008). Dual-task paradigms often increase sway, gait variability, or slowing, indicating that cognitive load competes with postural control processes (Herman et al., 2010; Yogev-Seligmann et al., 2008). Pauelsen et al. (2018) reported that a model containing centre-of-pressure measures from quiet-standing and limits-of-stability tasks explained 39% of the variance in Falls Efficacy Scale-International scores, while a model containing sensory and muscle-function measures explained 40%. Although this reflects falls-related concern rather than the occurrence of falls, it suggests that sensory, motor, and postural-control characteristics meaningfully contribute to how older adults perceive their capacity to perform activities without falling (Pauelsen et al., 2018). Executive-function deficits, particularly in attention, inhibition, and set-shifting, have also been linked with poorer mobility and increased falls-risk (Granacher et al., 2008; Herman et al., 2010; Mirelman et al., 2012; Yogev-Seligmann et al., 2008). This is important because everyday walking often requires balance control to occur alongside environmental monitoring, decision-making, and cognitive engagement. However, successful balance recovery also depends on whether centrally processed information can be translated into timely and effective muscular responses. This highlights the importance of the motor and musculoskeletal components of balance, particularly where age-related declines in strength and power may limit postural correction and recovery.

2.2.2.4 Motor and Musculoskeletal Components

Motor output translates integrated sensory information into stabilising actions through coordinated muscle activation patterns, commonly described through ankle, hip, and stepping strategies (Paillard, 2021; Peterka, 2018; Pollock et al., 2000; Winter, 1995). These strategies allow graded postural adjustments depending on the size and speed of a perturbation, with ankle strategies typically used for small disturbances and hip or stepping strategies used when larger corrections are required (Laughton et al., 2003; Pollock et al., 2000; Shumway-Cook & Woollacott, 2007; Winter, 1995). For balance recovery, the musculoskeletal system must generate an appropriate response before the centre-of-mass

moves beyond recoverable limits relative to the base of support (Maki & McIlroy, 2006; Pai et al., 1994).

With ageing, sarcopenia, reduced muscle power, impaired motor-unit recruitment, and a slower rate of force development can limit the capacity to generate rapid stabilising torque (Bellumori et al., 2013; Granacher et al., 2013; Granacher et al., 2008; Hester et al., 2021). This is particularly relevant for reactive stepping, because successful recovery from trips, slips, or external perturbations often requires a rapid and forceful response (Chang et al., 2005; Hester et al., 2021; Maki & McIlroy, 2006). Reductions in strength, muscle power, and intermuscular coordination can delay corrective reactions and reduce the effectiveness of stabilising responses (Granacher et al., 2008; Maki & McIlroy, 2006; Muehlbauer et al., 2015; Orr et al., 2008).

Reduced lower-limb strength and poorer neuromuscular function have been associated with impaired mobility, slower balance recovery, and greater falls-risk in older adults (Cho et al., 2012; Granacher et al., 2013; Wu et al., 2016). Functionally, reduced strength and muscle power may limit the ability to control body sway, take an effective recovery step, or control the transition between standing, walking, and turning (Horak, 2006; Maki & McIlroy, 2006; Shumway-Cook et al., 2000). Some older adults may rely more heavily on proximal or hip-dominant strategies when distal ankle control is reduced, although this compensation may be less efficient and may increase stiffness or fatigue during prolonged standing and walking (Granacher et al., 2013; Sturnieks et al., 2008; Winter, 1995).

Exercise interventions targeting neuromuscular performance, such as resistance, balance, and perturbation-based training, can improve strength, postural stability, and falls-related outcomes in older adults (Cadore et al., 2013; Granacher et al., 2013; Liu & Latham, 2009; Sherrington et al., 2019). Sherrington et al. (2019) reported that exercise reduced the rate of falls by 23% in community-dwelling older adults, based on 12,981 participants across 59 studies. Thus, the musculoskeletal system remains adaptable with appropriate stimuli; neuromuscular training may improve postural responses and reduce falls-risk by enhancing strength, power, coordination, and dynamic stability (Bird & Fell, 2014; Cadore et al., 2013; Granacher et al., 2013; Liu & Latham, 2009; Sherrington et al., 2019). Therefore, these sensory, cognitive, and neuromuscular determinants explain why balance reserve may

decline with age and why older adults may become more vulnerable to falls when task or environmental demands increase.

Overall, balance in older adults depends on the coordinated function of sensory input, sensory integration, cognitive regulation, and motor output (Horak, 2006; Sturnieks et al., 2008). Age-related changes in sensory, neuromuscular, and cognitive systems can reduce the reserve available for maintaining stability, particularly when task or environmental demands increase (Peterka, 2018; Sturnieks et al., 2008). However, balance is not expressed in only one way. The demands of quiet standing differ from those of walking, turning, transferring, or recovering from perturbations, as each task places different requirements on sensory integration, centre-of-mass control, stepping responses, and movement adaptation (Bohm et al., 2015; Horak, 2006; Maki & McIlroy, 2006). These differences provide the basis for distinguishing between static and dynamic balance in older adults.

2.3 Static and Dynamic Balance in Older Adults

The sensory, cognitive, neuromuscular, and musculoskeletal systems described above operate across a range of postural and movement tasks, but the demands placed on them differ depending on whether balance is maintained during relatively stationary standing or during movement (Horak, 2006; Pollock et al., 2000; Sturnieks et al., 2008). This distinction has shaped how balance is commonly conceptualised and assessed in older adults, with static and dynamic balance representing related but distinct expressions of postural control. Examining these two contexts separately helps clarify how each may contribute to understanding falls-risk in older adults.

2.3.1 Static Balance in Older Adults

Static balance refers to the ability to maintain a stable posture while the body's centre-of-mass remains within a relatively fixed base of support, such as during quiet standing or sitting upright (Pollock et al., 2000; Winter, 1995). In an upright stance, the body behaves like an inverted pendulum, pivoting primarily at the ankle joints. Even when a person appears to be standing still, the centre-of-mass is never entirely stationary. Instead, it moves subtly in the anterior-posterior and mediolateral directions, and the neuromuscular system continually generates small corrective responses to keep the body aligned over the feet

(Peterka, 2018; Winter, 1995). These small, continuous adjustments are illustrated in Figure 1, which depicts the characteristic sway pattern of quiet standing.

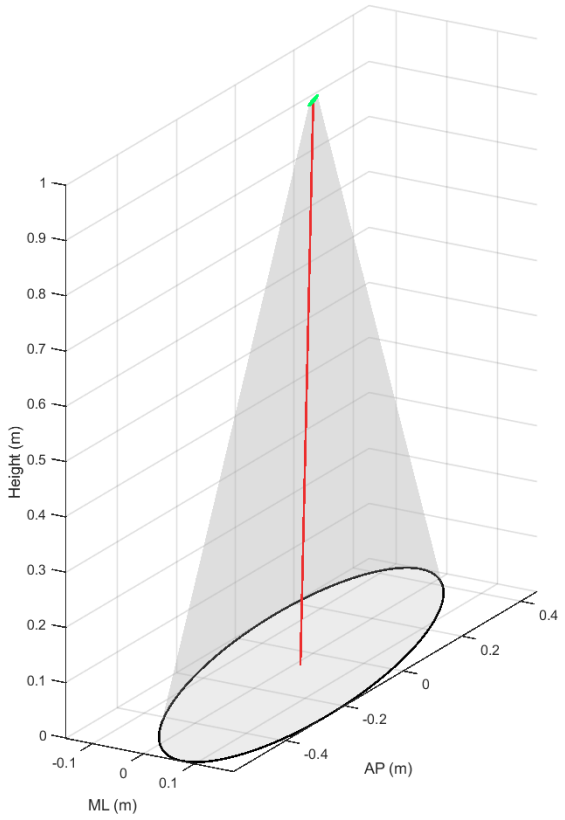


Figure 1. *Representative inverted-pendulum model of quiet standing from a participant classified as Low-Falls-Risk, illustrating anterior-posterior and mediolateral centre-of-pressure displacement relative to the estimated centre-of-mass position and body height*

In healthy younger adults, these automatic corrections are typically precise and well-coordinated, resulting in a relatively small and regular postural sway trajectory (Prieto et al., 1996; Winter, 1995). Studies consistently show that older adults use the same fundamental strategy for maintaining upright stance, but their sway pattern tends to be larger, more variable, and less regular, indicating that greater corrective effort is required to remain stable (Lord & Ward, 1994; Maki et al., 1994; Stelmach et al., 1989). Community-dwelling older adults without major pathology commonly demonstrate greater excursion of their centre-of-mass in both the anterior-posterior and mediolateral directions during quiet standing, although these movements generally remain within their functional base of support (Lord & Ward, 1994; Prieto et al., 1996).

Static balance becomes more demanding when the stance position reduces the base of support or limits sensory input for postural control. Conditions such as standing with feet close together, adopting a semi-tandem or tandem stance, or closing the eyes increase the difficulty of maintaining a steady posture (Manchester et al., 1989; Stelmach et al., 1989; Teasdale et al., 1991). Under these conditions, older adults typically show increased sway amplitude, higher sway velocity, and more irregular sway patterns than younger adults, reflecting the increased postural-control demands of maintaining upright stance when the base of support is reduced or sensory information is limited (Manchester et al., 1989; Teasdale et al., 1991). Research also shows that older adults are more likely to rely on supplementary stabilising behaviours during challenging stance tasks, including lightly touching nearby objects or increasing muscle co-contraction around the ankles and hips to enhance steadiness (Manchester et al., 1989; Stelmach et al., 1989).

Static balance also varies within the older adult population. Studies comparing older fallers and non-fallers indicate that individuals with a history of falls generally exhibit larger sway magnitudes, abrupt changes in body position, and greater difficulty maintaining demanding stances such as single-leg standing or narrow-base positions (Baloh et al., 1994; Maki et al., 1994). These differences are observed even among independently living older adults, suggesting that subtle impairments in static postural control may underlie increased vulnerability to balance loss in daily life (Lord & Ward, 1994; Maki et al., 1994). Collectively, these patterns show that older adults maintain static balance using the same core postural control mechanisms as younger adults, but the system functions closer to its stability limits and requires more continuous corrective adjustments to maintain an upright position (Peterka, 2018; Sturnieks et al., 2008).

However, static balance represents only one context in which postural control is required, as balance control is task-specific and depends on the interaction between the individual, the task, and the environment (Horak, 2006; Woollacott & Shumway-Cook, 2002). Many daily activities require stability while the body is moving, the centre-of-mass is continuously displaced, and the base of support changes from step-to-step (Maki & McIlroy, 2006; Winter, 1995). These demands highlight the importance of dynamic balance and gait, where postural

control must be integrated with forward progression, limb coordination, and adaptation to changing task demands.

2.3.2 Dynamic Balance and Gait in Older Adults

Dynamic balance refers to the ability to maintain stability while the body is in motion; it is required during everyday activities such as walking, turning, reaching, and completing transitional tasks like rising from a chair or stepping over an obstacle (Pollock et al., 2000; Sturnieks et al., 2008; Winter, 1995). Unlike quiet standing, where the centre-of-mass remains within a relatively stable base of support, dynamic balance involves continuously regulating the relationship between the centre-of-mass and the moving base of support as each foot alternates contact with the ground (Maki, 1997; Maki & McIlroy, 2006; Winter, 1995).

During walking, the centre-of-mass progresses anteriorly while undergoing rhythmic mediolateral and vertical excursions, and balance is maintained by coordinating this motion with the stance and swing phases of gait (Perry & Burnfield, 2010; Winter, 1995). The stance phase begins at initial contact, progresses through loading response and midstance as body weight is accepted over the supporting leg, and ends with terminal stance and toe-off as pressure shifts toward the forefoot to generate propulsion. The swing phase then occurs as the opposite leg moves forward in preparation for the next initial contact (Perry & Burnfield, 2010). Throughout these stages, the centre-of-pressure moves from the heel to the mid-foot and then to the forefoot, reflecting the natural pattern of foot loading during gait (Perry & Burnfield, 2010; Winter, 1995). For balance to be maintained, the stance leg and trunk must remain steady while the swing leg advances, ensuring that each step is placed accurately enough to guide the centre-of-mass back toward a stable position for the next cycle (Hausdorff, 2005; Maki, 1997; Maki & McIlroy, 2006; Sturnieks et al., 2008).

Figure 2 illustrates the major phases of the walking gait cycle, including alternating periods of stance, swing, and double support, as well as the corresponding loading pattern under each foot. Dynamic balance during gait, therefore, depends on coordinated foot loading, controlled movement of the centre-of-mass, and timely adjustments throughout each gait phase.

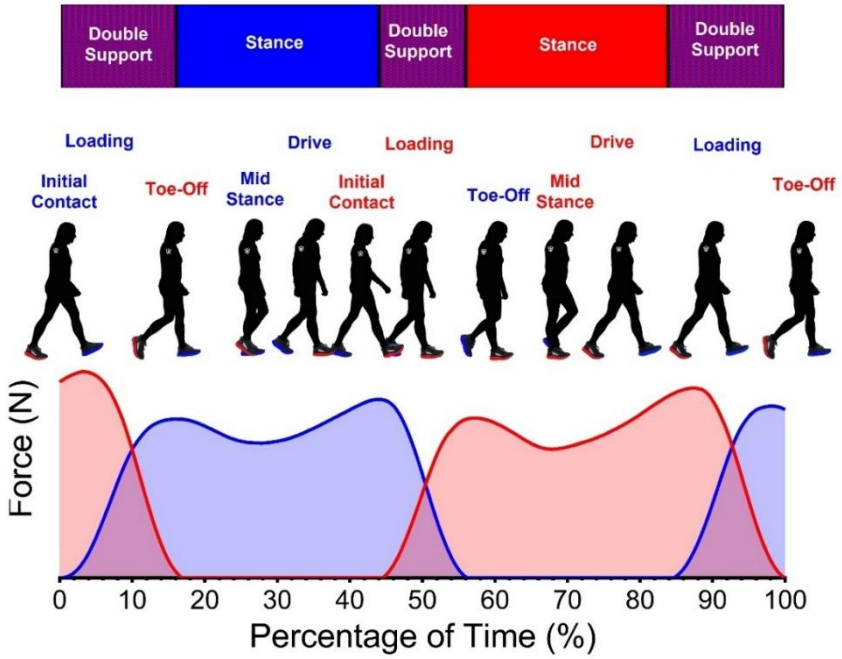


Figure 2. *Phases of the walking gait cycle and corresponding force-time pattern under each foot (Macdermid, 2025).*

In older adults, gait is commonly characterised by a slower walking speed, shorter step length, and a longer double-support period, meaning a greater proportion of each stride is spent with both feet on the ground (Laughton et al., 2003; Prince et al., 1997; Sadeghi et al., 2004; Shinichi & Tatsuya, 2015). These characteristics give gait a more cautious appearance and reduce the time spent in single-leg support, when balance must be maintained over a smaller and moving base of support (Maki, 1997; Perry & Burnfield, 2010). However, these changes should not be interpreted solely as deficits, as slower speed, shorter steps, and increased double-support time may also represent stabilising adaptations used to maintain balance during walking (Maki, 1997). Stride-to-stride variability is also typically greater, particularly in step timing and length, indicating less consistent control of leg placement from one step to the next (Fried et al., 2001; Hausdorff, 2005; Hausdorff et al., 2001; Maki, 1997). This variability is especially relevant to falls research because increased gait variability

has been identified as a predictor of future falls in community-dwelling older adults (Hausdorff et al., 2001; Maki, 1997).

Dynamic balance is also evident during tasks such as turning or navigating obstacles. During turning, older adults often rotate the head, trunk, and pelvis in more distinct segments rather than as a single coordinated unit, and their rotational movements tend to occur more slowly (Cromwell & Newton, 2004; Prince et al., 1997; Rogers et al., 2008; Sadeghi et al., 2004). When negotiating obstacles, older adults may adopt a range of compensatory strategies, including reducing walking speed, shortening step length, modifying support timing, and altering foot clearance (Chien et al., 2018; Pan et al., 2016; Raffegeau et al., 2022; Shin et al., 2015). These responses are not uniform, as some older adults increase foot clearance to create a greater safety margin, whereas others demonstrate reduced toe clearance, particularly during more demanding or unexpected obstacle-crossing conditions, which may increase the likelihood of contacting the obstacle (Maidan et al., 2018; Pan et al., 2016). Collectively, these movement patterns show that dynamic balance during turning and obstacle negotiation is maintained through ongoing adjustments in movement speed, step placement, support timing, and limb trajectory rather than through rigid postural control.

Transitional movements, such as rising from sitting or moving directly from sitting into walking, also place substantial demands on dynamic balance. Older adults may use compensatory strategies during these tasks, including altered arm use, foot positioning, and control of the centre-of-mass, to manage the transition from a stable seated position to upright stance and gait (Pai et al., 1994; van der Kruk et al., 2022). During sit-to-walk, the movement may be performed as a continuous transition in which gait begins immediately after rising, or as a more segmented sequence in which the individual first achieves upright standing before pausing and initiating the first step (Jones et al., 2018). In older adults, this greater separation between standing and gait initiation may reduce the complexity of the task by allowing postural stability to be established before forward progression begins (van der Kruk et al., 2022). These strategies illustrate how dynamic balance can be maintained by modifying the timing and organisation of transitional movements.

Research examining responses to minor slips, trips, or externally induced perturbations shows that older adults can adjust their stepping pattern to recover balance but tend to do so using slower, wider, or differently directed recovery steps, and sometimes require more

than one step to fully stabilise themselves (Gerards et al., 2021; Maki, 1997; Roos & Dingwell, 2013). When a disturbance occurs, they may widen the base of support by placing the recovery foot further laterally, alter the direction of the step, or take several successive steps to reposition the base of support relative to the centre-of-mass (Gerards et al., 2021; Maki, 1997; Pavol & Pai, 2007). Although adaptive stepping can still occur with repeated practice, the speed, consistency, and retention of these adaptations may be reduced in older adults, indicating that dynamic balance control remains flexible yet less streamlined in later life (Gerards et al., 2021; Maki, 1997; Van Wouwe et al., 2021).

Thus, age-related changes in gait speed, stride variability, turning, transitional movements, and recovery stepping illustrate how dynamic balance can become less consistent and increasingly reliant on compensatory strategies with age. These alterations are relevant to falls research because greater gait variability is associated with future falls, while cautious gait adaptations, transitional movement strategies, and recovery-stepping responses provide further insight into how older adults manage balance during everyday movement (Hausdorff et al., 2001; Maki, 1997; Sturnieks et al., 2008).

Overall, static and dynamic balance represent related but distinct expressions of postural control in older adults. Static balance provides insight into steadiness during relatively stationary positions, whereas dynamic balance reflects the ability to maintain stability during movement, including walking, transitions, turning, and responses to perturbations. Because these tasks place different demands on the postural-control system, understanding balance performance and its relationship to falls risk depends partly on the type of assessment used. Accordingly, the selection and interpretation of assessment methods are important when evaluating falls-risk, static balance, and dynamic balance in older adults.

2.4 Methods of Falls-Risk and Balance Assessment in Older Adults

Falls-risk and balance assessments can overlap, particularly when performance-based balance or mobility tests are used to help classify an individual's likelihood of falling (Lusardi et al., 2017; Montero-Odasso et al., 2022). Falls-risk assessment refers to the broader process of estimating or categorising an individual's likelihood of falling, and may draw on clinical history, self-reported risk factors, functional performance, physical assessment findings, or test-specific cut-off scores (Lusardi et al., 2017; Phelan et al., 2015; Stevens &

Phelan, 2013). Balance assessment evaluates postural or gait performance during specific tasks, such as standing, walking, turning, or responding to balance challenges (Horak, 2006; Mancini & Horak, 2010; Prieto et al., 1996). Some measures, therefore, provide task-specific information about balance performance while also contributing directly to falls-risk categorisation, particularly when balance or mobility scores are interpreted using falls-risk cut-off values (Lusardi et al., 2017; Montero-Odasso et al., 2022). This distinction is important because different assessment methods may capture different dimensions of falls-risk, from broad clinical screening to specific impairments in static or dynamic postural control. The following section, therefore, reviews the main approaches used to assess falls-risk and balance in older adults, including falls-risk screening tools, clinical balance assessments, and instrumented methods for evaluating static and dynamic balance.

2.4.1 Falls-Risk Screening and Assessment

Falls-risk screening provides an initial method to identify older adults who may be more vulnerable to future falls and may require further assessment. Because falls are influenced by multiple interacting factors, screening approaches commonly draw on information such as previous falls, unsteadiness, concern about falling, functional ability, medical history, medication use, cognition, frailty, and environmental risk (Lusardi et al., 2017; Montero-Odasso et al., 2022; Phelan et al., 2015). These tools are clinically useful because they can efficiently identify broad risk indicators, but their predictive performance varies, and they do not fully characterise the postural or gait mechanisms that may contribute to instability (Burns et al., 2022; Meekes et al., 2021; Montero-Odasso et al., 2022).

Falls-prevention guidance commonly begins with short case-finding questions before progressing to a more detailed assessment when risk is identified. The World Falls Guidelines recommend routinely asking older adults about falls, as falls are often not spontaneously reported during clinical encounters (Montero-Odasso et al., 2022). One example is the Three Key Questions, which ask about three clinically relevant indicators: whether the individual has fallen in the past year, feels unsteady when standing or walking, or is worried about falling (Montero-Odasso et al., 2022). These questions are important because previous falls are a strong predictor of future falls, while concern about falling has also been identified as an independent predictor of future falls in older adults (Deandrea et al., 2010; Ellmers et al., 2025; Montero-Odasso et al., 2022).

Brief question-based screening and questionnaire-based tools are widely used because they are quick to administer, require little or no equipment, and can be implemented in primary care, community, and public-health settings (Argyrou et al., 2023; Montero-Odasso et al., 2022; Phelan et al., 2015; Stevens & Phelan, 2013). They also capture information that may not be visible during a short physical test, such as falls history, perceived unsteadiness, concern about falling, medication use, vision problems, and functional difficulty (Argyrou et al., 2023; Burns et al., 2022; Phelan et al., 2015). A systematic review of question-based falls-risk tools in community-dwelling older adults identified considerable variation in the number of items included, the risk factors assessed, and the intended screening context (Argyrou et al., 2023). This variation reflects the broad range of factors that contribute to falls but also means that tools differ in what they capture and how well they identify future falls. Self-report can also be influenced by recall, insight, symptom interpretation, and willingness to disclose difficulty (Cummings et al., 1988; Ganz et al., 2005; McInnes & Askie, 2004; Yardley et al., 2006; Zecevic et al., 2006). Falls may be forgotten or under-reported, particularly when recall periods are long or when the fall did not result in injury (Cummings et al., 1988; Ganz et al., 2005). Older adults may also differ in how they define a fall and may minimise or withhold falls-related concerns because they are worried about appearing frail, being perceived as dependent or incapable, or losing autonomy and independence (Gardiner et al., 2017; Yardley et al., 2006; Zecevic et al., 2006). Embarrassment, stigma, and a desire to maintain an identity as independent and capable may therefore influence whether falls, unsteadiness, or concern about falling are fully disclosed (Gardiner et al., 2017; Yardley et al., 2006). Therefore, question-based tools are useful for initial risk identification, but they should be interpreted alongside other sources of clinical and performance-based information where possible.

The Stopping Elderly Accidents, Deaths, and Injuries initiative, commonly known as STEADI, is a structured screening, assessment, and intervention framework developed by the Centres for Disease Control and Prevention (CDC) to support falls-risk identification and prevention in clinical practice (CDC, 2023; Phelan et al., 2015; Stevens & Phelan, 2013). The STEADI algorithm lists two screening options for community-dwelling adults aged 65 years and over: the Three Key Questions and the Stay Independent questionnaire (CDC, 2023; Montero-Odasso et al., 2022). The Three Key Questions ask whether the individual feels unsteady

when standing or walking, worries about falling, or has fallen in the past year, while the Stay Independent questionnaire includes 12 items related to falls history, unsteadiness, worry about falling, medication use, strength, vision, sensation, toileting, and mood (CDC, 2023; Phelan et al., 2015; Stevens & Phelan, 2013). A strength of STEADI is that it provides a clear pathway from initial screening to assessment of modifiable risk factors and intervention planning for individuals identified as being at risk. However, this staged structure also means that further assessment depends on the initial screening result. Older adults who do not report falls, unsteadiness, or concern about falling may be classified as not at risk and therefore may not progress to more detailed assessment, even if subtle balance or gait impairments are present.

Evidence supports the clinical relevance of STEADI as a structured falls-prevention pathway, but findings regarding its predictive performance are mixed. Lohman et al. (2017) operationalised an adapted STEADI algorithm in a nationally representative sample of adults aged 65 years and over and found that those classified as having moderate or high falls-risk had greater odds of subsequent falls than those classified as low risk. Kannoth et al. (2021) adapted the STEADI Key Questions for a large prospective cohort of older adult drivers and found that baseline classification was associated with future falls over two years. Older adults classified as at risk had higher odds of both any fall and multiple falls, although the adapted questions demonstrated only fair predictive ability. However, other studies have raised concerns about the accuracy and consistency of STEADI-based classification. Nithman and Vincenzo (2019) found that STEADI sensitivity differed between community-dwelling and retirement-facility samples and reported false-negative cases among individuals classified as low risk. They suggested that STEADI cut-off scores, functional-test selection, and the weighting of self-report items may require further evaluation to improve discriminative and predictive validity. Similarly, Burns et al. (2022) found that falls-risk screening tools, including the Three Key Questions and Stay Independent questionnaire, showed variable sensitivity and specificity for predicting falls over 12 months. Lin et al. (2022) likewise concluded that STEADI's predictive validity varies across studies and may depend on how the algorithm is operationalised, the population assessed, and the screening components included. Collectively, these findings suggest that STEADI provides a useful structure for identifying falls-risk and guiding further assessment, but its classifications

should be interpreted as one component of a broader falls-risk profile rather than as a complete measure of underlying balance, gait, or physiological vulnerability.

Other multidomain falls-risk tools have been developed for different settings and purposes. The Falls-Risk for Older People in the Community tool, or FROP-Com, was designed to assess multiple falls-risk factors in community-dwelling older adults and includes areas such as falls history, medications, medical conditions, sensory loss, cognition, continence, nutrition, environment, functional behaviour, balance, gait, and physical activity (Mascarenhas et al., 2019; National Ageing Research Institute, 2020). By assessing a comprehensive profile of risk factors, FROP-Com can aid in developing targeted intervention plans. However, Mascarenhas et al. (2019) reported that a high-risk FROP-Com classification had limited sensitivity for predicting future falls in older adults who had presented to an emergency department following a fall and were subsequently discharged home. The authors suggested that in this high-risk population, the tool may be more useful for identifying individual risk factors and guiding targeted intervention than for accurately classifying who will or will not fall (Mascarenhas et al., 2019). This illustrates an important distinction in falls-risk assessment: some tools may be more useful for identifying modifiable contributors to risk than for predicting falls with high accuracy.

The Falls-Risk Assessment Tool, or FRAT, is another structured tool that combines initial falls-risk screening with broader risk-factor identification and intervention planning (Stapleton et al., 2009). Its staged structure makes it useful for linking screening to risk-factor identification and intervention planning, particularly in subacute and residential-care settings (Stapleton et al., 2009). However, as with other tools, its relevance depends on the population and setting in which it is used. Several other tools, including STRATIFY, the Morse Fall Scale, and the Hendrich II Fall-Risk Model, are widely discussed in the falls literature, particularly in acute-care or inpatient settings (Aranda-Gallardo et al., 2013; Strini et al., 2021). These tools are important in hospital contexts because they support the structured identification of patients who may be at increased risk of inpatient falls (Aranda-Gallardo et al., 2013; Strini et al., 2021). However, their predictive performance varies across populations and clinical environments, meaning that findings from inpatient settings may not translate directly to independently living older adults in the community (Aranda-Gallardo et al., 2013; Strini et al., 2021). Tool selection should therefore consider the

population, environment, available time, and whether the purpose is initial screening, risk stratification, or intervention planning.

Psychological screening tools may also contribute useful information, particularly when the aim is to understand concern about falling, fear-related avoidance, or balance confidence (Delbaere et al., 2004; Hadjistavropoulos et al., 2011; Zijlstra et al., 2007). The Falls Efficacy Scale-International and Short Falls Efficacy Scale-International assess concern about falling during everyday physical and social activities, while the Activities-Specific Balance Confidence Scale measures confidence in performing activities without losing balance or becoming unsteady (Powell & Myers, 1995; Yardley et al., 2005). These tools are not comprehensive falls-risk assessments, but they provide valuable insight into psychological and behavioural dimensions of risk, including perceived vulnerability, avoidance, and activity restriction (Delbaere et al., 2004; Hadjistavropoulos et al., 2011; Zijlstra et al., 2007). Therefore, psychological measures can complement broader falls-risk screening by capturing confidence-related and behavioural factors that may not be fully reflected in physical performance measures alone (Ellmers et al., 2025; Hadjistavropoulos et al., 2011; Yardley et al., 2005).

Frailty and functional screening tools may also contribute to falls-risk assessment because they identify broader physiological vulnerability. Frailty reflects reduced physiological reserve and is associated with adverse outcomes, including disability, mobility limitation, and increased vulnerability to stressors (Cesari et al., 2017; Fried et al., 2001). The Kihon Checklist is one example of a multidomain frailty and functional screening tool, incorporating areas such as instrumental activities of daily living, physical activity, nutrition, oral function, socialisation, cognition, and mood (Watanabe et al., 2022). Watanabe et al. (2022) reported that the Kihon Checklist demonstrated sufficient diagnostic accuracy for identifying frailty in older adults when compared with a physical frailty phenotype reference standard. Frailty screens are relevant because frailty and falls-risk share several contributing factors, including weakness, slowed gait, low activity, cognitive impairment, and functional decline (Cesari et al., 2017; Fried et al., 2001; Montero-Odasso et al., 2022). However, frailty and falls-risk should not be treated as identical constructs, because frailty screening identifies broader physiological vulnerability, whereas falls-risk assessment focuses more

specifically on the likelihood of future falls and the modifiable contributors to that risk (Cesari et al., 2017; Fried et al., 2001; Montero-Odasso et al., 2022).

Overall, falls-risk screening tools provide clinically useful information about future vulnerability, but their strengths and limitations differ according to the tool, population, and setting. Furthermore, these tools do not directly quantify how an individual controls posture or movement during specific tasks. Clinical balance and mobility tests complement this screening information by assessing performance during standing, walking, turning, and transitional movements.

2.4.2 Clinical Tests of Static and Dynamic Balance

Clinical balance assessments are widely used in geriatric, community, and rehabilitation settings because they are simple to administer, require minimal equipment, and provide functional insight into how well an individual maintains stability during everyday activities (Berg et al., 1992; Podsiadlo & Richardson, 1991). These tests evaluate the ability to control posture while standing still, transitioning between positions, and moving through space (Berg et al., 1992; Podsiadlo & Richardson, 1991). Although they do not quantify the underlying mechanisms of postural control, they provide practical information on stability in tasks that older adults routinely perform (Mancini & Horak, 2010).

One of the most commonly used functional assessments is the Timed Up and Go (TUG), developed by Podsiadlo and Richardson (1991). The TUG involves standing up from a standard chair, walking three metres, turning, walking back, and sitting down. The total time required reflects the integration of multiple balance components, including transitional control, gait stability, turning ability, and coordination of lower-limb movements (HQSC, 2020; Podsiadlo & Richardson, 1991; Shumway-Cook et al., 2000). It is widely used in clinical and community screening, and in Aotearoa New Zealand national falls-prevention guidance, such as the HQSC's Falls-risk assessment: a multifactorial approach document, TUG is recommended as one of the key mobility measures (HQSC, 2020). Evidence for the TUG's reliability and validity comes from the original validation study, which reported strong inter-rater and intra-rater reliability and significant associations with other functional measures, including the Berg Balance Scale, gait speed, and activities of daily living (Podsiadlo & Richardson, 1991). Shumway-Cook et al. (2000) also showed that TUG performance,

particularly under dual-task conditions, could help distinguish older adults with and without a history of recurrent falls. However, while these findings support the TUG as a practical measure of functional mobility, later evidence suggests that its ability to predict falls is limited when used in isolation. Barry et al. (2014) reported that the TUG has limited ability to predict falls in community-dwelling older adults and should not be used alone to identify individuals at high fall-risk. More recently, Nakanowatari, Hoshi, Sone, et al. (2024) found that TUG time did not differ significantly between older adult fallers and non-fallers, whereas peak force asymmetry measured using in-shoe sensors was significantly greater in fallers. This suggests that TUG completion time may reflect broad functional mobility but may not capture subtle biomechanical differences in gait loading or asymmetry that are associated with fall history in higher-functioning older adults. For this reason, TUG results are typically interpreted alongside other balance assessments.

Broader aspects of functional balance are captured by the Berg Balance Scale (BBS), a 14-item assessment that includes static tasks, such as quiet standing, feet together, and single-leg stance; transitional tasks, such as sit-to-stand and reaching; and movement-related tasks, such as turning, stepping, and retrieving objects from the floor (Berg et al., 1992). The BBS provides a graded profile of balance performance across tasks of increasing difficulty, and scores have been associated with fall history and fall-risk in older adults (Berg et al., 1992; Muir et al., 2008). However, limitations include ceiling effects in higher-functioning older adults and reduced sensitivity to small changes in balance performance (Mancini & Horak, 2010). These limitations suggest that although the BBS is useful for identifying moderate impairments, it may not detect subtle deficits that appear only in more demanding or instrumented tests.

Static balance is also commonly assessed using the Four-Stage Balance Test, a brief screening tool commonly used in community falls-risk assessment. Individuals attempt to hold four progressively narrower stances: feet together, semi-tandem, tandem, and single-leg standing, each for up to 10 seconds (CDC, 2017; Rossiter-Fornoff et al., 1995). These positions reduce the width of the base of support, making balance control more challenging. Difficulty maintaining the tandem stance for 10 seconds is commonly used as an indicator of increased fall-risk in older adults (CDC, 2017). Although simple, the Four-Stage Balance Test

provides practical information about static balance performance and can help identify individuals who may require further gait, balance, or falls-risk assessment.

Other clinical measures contribute additional perspectives on balance. The Functional Reach Test, introduced by Duncan et al. (1990), quantifies anticipatory postural control by measuring how far an individual can reach forward without stepping. Reduced reach distance is common in older adults and is associated with higher fall-risk and poorer functional balance (Duncan et al., 1990; Maki et al., 1994). Additionally, the 30-Second Chair Stand Test, widely used in community screening, measures how many sit-to-stand repetitions an individual can complete within 30 seconds (Jones et al., 1999). Although primarily an indicator of lower-limb strength and functional mobility, it also reflects components of dynamic balance, as successful performance requires forward weight shifting and stable re-alignment during each repetition (Whitney et al., 2005).

Clinical walking assessments also appear in the literature. The Tandem Walk Test, for example, requires individuals to walk heel-to-toe along a straight line. Performance reflects dynamic balance and mediolateral stability, as this narrow-base gait challenges the ability to control side-to-side motion (Robertson & Gregory, 2017). Difficulty performing tandem walking is common in adults with impaired vestibular or somatosensory function and is associated with increased fall-risk in older adults (Kristinsdottir et al., 2001; Manchester et al., 1989). Variations of this assessment can include eyes-closed conditions or foam-surface walking to further challenge sensory integration (Manchester et al., 1989; Teasdale et al., 1991).

Another widely used gait assessment is the 10-Metre Walk Test (10MWT), which measures comfortable or fast walking speed over a short, standardised distance (Peters et al., 2013; Rahman & Alagappan, 2019). Walking speed is considered a powerful indicator of overall mobility, physiological reserve, and the ability to walk safely and independently in everyday environments, making the 10MWT a practical measure of functional walking ability in older adults (Boripuntakul et al., 2024; Middleton et al., 2015). Slower gait speed in older adults is consistently associated with higher falls-risk, reduced functional independence, and increased likelihood of future mobility limitations (Barak et al., 2006; Barbosa & Souza, 2022; Fernandez et al., 2019). The 10MWT is straightforward to administer, demonstrates strong reliability for measuring walking speed in older adults, and is responsive to age-related

changes in mobility (Peters et al., 2013; Rahman & Alagappan, 2019). However, as with other clinical walking tests, gait speed alone provides limited information about the biomechanical strategies underlying walking performance, such as asymmetry, loading patterns, or compensatory movement strategies. For example, Nakanowatari, Hoshi, Sone, et al. (2024) found that 10MWT speed did not differ significantly between older adult fallers and non-fallers, whereas peak force asymmetry measured using in-shoe sensors was significantly greater in fallers.

Collectively, these clinical tests offer valuable insight into how older adults maintain or regain stability during everyday movements. They are accessible, inexpensive, and straightforward to administer across community, clinical, and research settings, which is why they remain core components of geriatric balance assessment (Berg et al., 1992; Podsiadlo & Richardson, 1991; Rahman & Alagappan, 2019). However, despite their strengths, several limitations are well documented in the literature. First, clinical tests provide limited information about the underlying mechanisms of postural control. They cannot quantify centre-of-pressure behaviour, sway variability, sensory weighting, or neuromuscular response characteristics, meaning they may overlook subtle deficits that instrumented tools can detect (Mancini & Horak, 2010; Prieto et al., 1996; Winter, 1995). Second, performance on clinical tests can be influenced by factors unrelated to balance, including musculoskeletal pain, reduced lower-limb strength, cardiopulmonary limitations, or cognitive impairments, all of which may affect test outcomes independently of postural stability (Lord & Ward, 1994; Maki et al., 1994). Third, several clinical tests demonstrate ceiling or floor effects. The Berg Balance Scale, for instance, often fails to detect small declines in higher-functioning older adults, whereas single-leg stance tests may be too difficult for lower-functioning individuals, thereby reducing sensitivity across diverse populations (Maki et al., 1994; Mancini & Horak, 2010).

For these reasons, clinical assessments are most useful when interpreted alongside other sources of falls-risk information rather than used in isolation. Clinical tests, therefore, remain important because they reflect the practical demands of daily life and provide an efficient method of screening for mobility limitations and potential fall-risk (Berg et al., 1992; Muir et al., 2008; Podsiadlo & Richardson, 1991). Nevertheless, both clinical and self-reported measures have limited capacity to quantify the underlying biomechanical and physiological

mechanisms of balance control. Instrumented assessments address this limitation by detecting changes in centre-of-pressure behaviour, postural sway, gait loading, and asymmetry that may not be visible during functional tests. Instrumented methods can therefore provide a more detailed assessment of static balance.

2.4.3 Instrumented Measures of Static Balance

Force platforms include both laboratory-grade and portable force-plate systems and represent one of the most established instrumented approaches in balance research (O'Connor et al., 2016; Piirtola & Era, 2006; Prieto et al., 1996; Winter, 1995). Laboratory-grade systems are commonly used as fixed platforms in biomechanics laboratories and allow detailed measurement of ground-reaction forces, moments, and centre-of-pressure trajectories, whereas portable systems provide a more accessible method of collecting centre-of-pressure-based balance measures outside specialised laboratory settings (O'Connor et al., 2016; Pajala et al., 2008; Piirtola & Era, 2006; Prieto et al., 1996; Winter, 1995). These devices contain force transducers that measure ground-reaction forces and moments beneath the feet, allowing calculation of centre-of-pressure behaviour during standing balance tasks (Pajala et al., 2008; Piirtola & Era, 2006; Prieto et al., 1996; Winter, 1995). Laboratory-grade force platforms can measure ground-reaction forces in the vertical, anterior-posterior, and mediolateral directions, and depending on the system and protocol, provide detailed force and centre-of-pressure data for biomechanical analysis (Prieto et al., 1996; Winter, 1995).

Force platforms are also well-suited to exploring how sensory inputs contribute to balance. Researchers commonly manipulate visual input, such as comparing eyes-open and eyes-closed conditions, to examine the contribution of vision to postural control (Manchester et al., 1989; Teasdale et al., 1991). Somatosensory input can also be challenged by changing the support surface, such as comparing firm-surface standing with more compliant surface conditions, while stance width can be altered to increase postural-control demands (Manchester et al., 1989; Pajala et al., 2008; Piirtola & Era, 2006; Teasdale et al., 1991). In this context, "less reliable" sensory information refers to conditions in which one sensory system provides reduced or less precise information for maintaining upright posture. For example, closing the eyes removes visual input, while standing on a compliant surface reduces the precision of somatosensory feedback from the feet and ankles (Manchester et

al., 1989; Teasdale et al., 1991). Older adults typically show larger increases in sway when visual or somatosensory information is reduced, indicating reduced ability to maintain stability when sensory conditions become more challenging (Kristinsdottir et al., 2001; Manchester et al., 1989; Teasdale et al., 1991). For example, Manchester et al. (1989) showed that older adults had greater difficulty maintaining balance when visual, vestibular, or somatosensory information was disrupted, while Teasdale et al. (1991) demonstrated age-related differences in the integration of visual information during standing balance tasks. These findings suggest that force-platform testing can provide insight into how ageing affects multisensory integration and postural-control strategies (Kristinsdottir et al., 2001; Manchester et al., 1989; Teasdale et al., 1991).

In addition to laboratory-grade force platforms, portable force-plate systems have become increasingly used in community, clinical, and applied research settings because they provide a more accessible method of quantifying postural sway (Goble et al., 2018; O'Connor et al., 2016; Richmond et al., 2018). These systems measure ground-reaction forces beneath the feet and use the resulting force and moment signals to calculate centre-of-pressure coordinates during standing balance tasks (Winter, 1995). From the centre-of-pressure trajectory, processed outcomes such as path length, mean velocity, directional displacement, root-mean-square displacement, range, and confidence-ellipse area can be calculated to describe different spatial and temporal features of postural sway (Prieto et al., 1996; Quijoux et al., 2021). Although portable and laboratory-grade force platforms use the same general biomechanical principles to quantify centre-of-pressure behaviour, systems may differ in their sensor configuration, sampling frequency, signal processing, and the range of centre-of-pressure variables reported by their accompanying software (Quijoux et al., 2021; Richmond et al., 2018). Commercial portable systems may therefore present a predefined set of automatically processed balance outcomes through standardised software interfaces, simplifying data collection and interpretation for applied use (Goble et al., 2019; Richmond et al., 2018). Their portability, calibrated hardware, standardised testing protocols, and automated analysis make them useful for obtaining repeatable centre-of-pressure measurements outside specialised laboratory environments (Goble & Baweja, 2018; Goble et al., 2019; O'Connor et al., 2016).

Research evaluating portable balance force-plate systems supports their validity, accuracy, precision, and agreement with laboratory-grade or known force-application measures. O'Connor et al. (2016) compared two BTrackS Balance Plates with a laboratory-grade force platform and reported almost perfect agreement for centre-of-pressure measurement, with intraclass correlation coefficients greater than 0.999, error magnitude below 1%, and regression residuals below 0.2 mm. Goble et al. (2018) extended this validation using a point-of-application approach and reported excellent agreement between applied force locations and BTrackS-measured centre-of-pressure values, with intraclass correlation coefficients greater than 0.999, average percentage error below 1%, and average residual error below 0.1 cm. Richmond et al. (2018) also reported strong agreement between BTrackS and laboratory force-platform outcomes, with 35 of 38 postural-stability variables showing correlations of 0.98 or greater across rigid and compliant surface conditions. These findings support the use of BTrackS as a valid portable tool for quantifying postural sway, particularly when the aim is to obtain practical centre-of-pressure-based measures rather than full biomechanical force-platform data (Goble et al., 2018; O'Connor et al., 2016; Richmond et al., 2018). However, BTrackS has limitations. Compared with laboratory-grade force platforms, it provides less comprehensive biomechanical information because it is primarily designed to quantify centre-of-pressure-based postural sway rather than full three-dimensional ground-reaction forces, force-time variables, impulse, or loading-rate characteristics (Goble et al., 2018; O'Connor et al., 2016; Prieto et al., 1996; Richmond et al., 2018; Winter, 1995). Furthermore, BTrackS outcomes should be interpreted in relation to the testing protocol, as centre-of-pressure measures can vary according to stance position, support surface, visual input, and footwear (Lord & Bashford, 1996; Manchester et al., 1989; Prieto et al., 1996; Teasdale et al., 1991). It also cannot identify the specific sensory, muscular, or joint-level strategies used to maintain balance, meaning that BTrackS outcomes are best interpreted as practical indicators of static postural sway rather than complete descriptions of the mechanisms underlying postural control (Mancini & Horak, 2010).

Inertial measurement units (IMUs) and accelerometers have also emerged as valuable tools for static balance assessment. When worn on the trunk, pelvis, or lower back, these sensors measure small oscillations in acceleration or angular velocity that occur during quiet standing (Baker et al., 2021; Ghislieri et al., 2019). From these data, indices such as root-

mean-square acceleration, jerk, sway regularity, and frequency-domain characteristics can be derived to quantify postural stability (Baker et al., 2021; Ghislieri et al., 2019; Mancini & Horak, 2010). Evidence from systematic reviews suggests that wearable inertial sensors show moderate concurrent validity and moderate-to-excellent reliability for balance assessment, although results vary across sensor placement, testing condition, population, and outcome variable (Baker et al., 2021; Ghislieri et al., 2019). Therefore, rather than being treated as direct replacements for force platforms, IMUs are best viewed as complementary tools that estimate body-segment sway and movement behaviour using acceleration and angular-velocity signals (Baker et al., 2021; Ghislieri et al., 2019). IMUs can detect age-related and condition-related differences in balance, including changes in mediolateral sway amplitude and altered postural-sway characteristics under different testing conditions (Baker et al., 2021; Ghislieri et al., 2019). Their portability allows IMUs to be used not only in laboratory environments but also in clinics, homes, and residential care facilities, making repeated monitoring feasible and potentially improving the ecological relevance of balance assessment (Baker et al., 2021; Ghislieri et al., 2019; Vavasour et al., 2021). However, IMU-based balance assessment also has limitations. Outcomes can be influenced by sensor placement, attachment firmness, calibration, filtering choices, and analysis algorithms, and IMUs do not directly measure centre-of-pressure or ground-reaction forces (Baker et al., 2021; Ghislieri et al., 2019). Accordingly, IMUs provide valuable information about body-segment motion during balance tasks, but their outputs should be interpreted alongside the specific testing protocol and validation evidence for the variables being used (Baker et al., 2021; Ghislieri et al., 2019; Vavasour et al., 2021).

Instrumented static assessments provide objective information about postural steadiness during standing; however, many falls occur during movement-based activities such as walking and transferring between positions (Maki, 1997; Robinovitch et al., 2013). Instrumented dynamic balance and gait assessments, therefore extend evaluation beyond quiet standing by examining how older adults regulate stability during walking and functional mobility.

2.4.4 Instrumented Measures of Dynamic Balance and Gait

One of the most widely used approaches involves wearable IMUs. Unlike their use in static balance, where they quantify small postural oscillations, IMUs applied during gait capture dynamic movement characteristics, including stride length, gait speed, trunk acceleration patterns, and step-to-step variability (Baker et al., 2021; Vavasour et al., 2021; Weiss et al., 2013; Yin et al., 2025). Common sensor placement sites include the lower back, pelvis, shins, or feet, which allows detection of how the trunk moves over the base of support and how consistently the legs are swung from step to step (Baker et al., 2021; O'Brien et al., 2019; Yin et al., 2025). IMUs are particularly informative for identifying subtle gait irregularities, such as increased mediolateral trunk acceleration or irregular timing between strides, which are markers of reduced dynamic stability in older adults (Hausdorff et al., 2001; Weiss et al., 2013). IMUs are lightweight and require minimal set-up, they allow gait to be measured not only in laboratory walkways but also in corridors, clinics, and free-living environments, thereby improving ecological validity (Baker et al., 2021; Vavasour et al., 2021; Yin et al., 2025). However, IMU-based gait assessment also has limitations. Data quality can be influenced by sensor placement, attachment movement, calibration, filtering decisions, and the algorithms used to identify gait events or derive spatiotemporal variables (Baker et al., 2021; Yin et al., 2025). In addition, IMUs estimate body-segment motion rather than directly measuring ground-reaction forces or plantar loading, meaning that their outputs should be interpreted in relation to the specific sensor location, protocol, and validation evidence for the variables being used (Baker et al., 2021; Yin et al., 2025).

Another important category of tools used to assess dynamic balance is instrumented pressure-sensitive insoles, which measure plantar loading during gait (Razak et al., 2012). These systems have been evaluated against force-platform measurements and can provide valid and reliable estimates of vertical force-based variables during walking and running, although accuracy can vary according to walking or running condition, sampling frequency, footwear, calibration, and the specific variable being analysed (Mulligan & Milner, 2025; Renner et al., 2019; Seiberl et al., 2018). For example, Renner et al. (2019) reported that the Loadsol® system demonstrated generally good-to-excellent validity for peak vertical force across walking and running conditions, whereas loading-rate and impulse estimates showed greater variability and improved when higher-frequency insoles were used. These devices

quantify plantar loading patterns throughout the stance phase of gait, providing real-time measurements of vertical force, loading rate, contact time, impulse, and regional loading under the rearfoot, midfoot, and forefoot (Mulligan & Milner, 2025; Nakanowatari, Hoshi, Asao, et al., 2024; Nakanowatari, Hoshi, Sone, et al., 2024; Renner et al., 2019). This allows researchers to examine how weight is accepted and transferred across the foot during initial contact, midstance, and push-off.

The most relevant in-shoe sensor variables depend on the research question. In older-adult falls research, peak force asymmetry appears particularly important because Nakanowatari, Hoshi, Sone, et al. (2024) found that peak force asymmetry was significantly greater in fallers than non-fallers, whereas 10MWT speed and TUG time did not differ significantly between groups. In contrast, in prospective frailty research, Nakanowatari, Hoshi, Asao, et al. (2024) found that net impulse asymmetry was independently associated with the development of frailty over 10 months, with an adjusted odds ratio of 1.269 for each 1% increase. Therefore, peak force, impulse, loading rate, contact time, and left-right asymmetry are useful variables, but their relevance may differ depending on whether the focus is on fall history, frailty development, or general gait loading. These findings suggest that in-shoe sensors can detect gait-loading features that may not be captured by clinical timing measures alone (Nakanowatari, Hoshi, Asao, et al., 2024; Nakanowatari, Hoshi, Sone, et al., 2024). However, in-shoe sensor outputs should be interpreted in the context of the walking task and measurement conditions, as values may be influenced by shoe fit, cushioning, calibration, walking speed, and sampling characteristics (Razak et al., 2012; Renner et al., 2019; Seiberl et al., 2018). Instrumented insoles also primarily quantify vertical plantar loading and therefore do not provide the full three-dimensional kinetic information available from laboratory force-platform systems (Mulligan & Milner, 2025; Renner et al., 2019; Seiberl et al., 2018).

Additional laboratory-based gait assessment tools, such as three-dimensional motion-capture systems, instrumented walkways, and instrumented treadmills, are also frequently used in research. Motion-capture systems track joint and segment movement to provide detailed information on step width, foot trajectory, trunk rotation, and leg coordination (Cieślik et al., 2024; Huang et al., 2022; Prince et al., 1997; Sadeghi et al., 2004; Scataglini et al., 2024). Instrumented walkways contain embedded pressure or sensor arrays that record

temporospatial parameters, including step time, stance duration, double-support time, and stride variability, and have demonstrated strong reliability for quantifying gait parameters in older adults (Menz et al., 2004; Webster et al., 2005). Instrumented treadmills can also provide spatiotemporal and kinetic gait information over many consecutive steps, making them useful for examining gait variability, responses to controlled walking speeds, fatigue effects, or changes across repeated gait cycles (Song & Hidler, 2008; Watt et al., 2010). However, treadmill walking is not always directly equivalent to overground walking. Belt movement, fixed speed, handrail use, acclimatisation, and altered visual flow can influence gait mechanics, and studies in older adults have reported differences in cadence, stride length, stride time, and joint kinetics between treadmill and overground conditions (Song & Hidler, 2008; Watt et al., 2010). Studies indicate that increased step-to-step variability and longer double-support time are common in older adults and are associated with reduced dynamic stability and higher fall-risk (Hausdorff et al., 2001; Maki, 1997).

Thus, a wide range of instrumented methods allows researchers to examine dynamic balance from multiple perspectives, capturing how older adults regulate movement, manage foot-ground interactions, and maintain stability during walking (Hausdorff et al., 2001; Maki, 1997; Vavasour et al., 2021; Yin et al., 2025). These measures may reveal subtle differences in gait performance that are not captured by conventional clinical assessments, although their interpretation depends on the capabilities and limitations of each system (Nakanowatari, Hoshi, Asao, et al., 2024; Nakanowatari, Hoshi, Sone, et al., 2024; Yin et al., 2025). Instrumented dynamic balance and gait assessments provide objective information about how stability is regulated during movement, but sensory-specific testing can add further insight into the systems contributing to impaired postural control. Sensory-specific assessments can therefore help identify how visual, vestibular, and somatosensory function contribute to balance and falls-risk.

2.4.5 Sensory-Specific Assessments

Balance depends on the coordinated contribution of the visual, somatosensory, and vestibular systems; therefore, several assessment methods have been developed to alter, challenge, or selectively reduce these sensory inputs (Assländer & Peterka, 2014; Peterka, 2018; Sturnieks et al., 2008). These approaches help identify how strongly older adults rely on specific sensory inputs, how effectively sensory reweighting occurs when information

becomes less reliable, and where age-related sensory decline may contribute to instability (Assländer & Peterka, 2014; Manchester et al., 1989; Peterka, 2018).

Somatosensory function is commonly assessed using vibration-based tests, which quantify sensory acuity at sites such as the ankle, Achilles tendon, or tibialis anterior or through tests of lower-limb proprioceptive and vibration sensitivity (Horak, 2006; Johansson & Magnusson, 1991; Lord et al., 1991; Skinner et al., 1984). In these assessments, a vibration source, such as a tuning fork, biothesiometer, or similar device, is applied to a bony prominence or tendon, and the individual is asked to report when the vibration is first perceived and when it is no longer detectable or to identify the threshold at which vibration can be detected (Lord et al., 1991; Skinner et al., 1984). The duration of perceived vibration or the minimum amplitude required for detection provides an index of vibration sensitivity, with higher vibration thresholds indicating reduced somatosensory function (Lord et al., 1991; Skinner et al., 1984). Age-related reductions in proprioceptive and vibration sensitivity have been reported in older adults and are relevant to balance control because reduced lower-limb sensory information can impair the ability to detect body movement relative to the support surface (Lord et al., 1991; Skinner et al., 1984; Sturnieks et al., 2008). These sensory alterations have been associated with increased postural sway and poorer stability during quiet standing (Kristinsdottir et al., 2001; Manchester et al., 1989).

Some experimental protocols further introduce vibratory perturbations while individuals stand on a force platform to examine how the balance system responds when proprioceptive information is altered (Fransson et al., 1998; Fransson et al., 2001; Johansson et al., 2001; Johansson & Magnusson, 1991). In these paradigms, small mechanical vibrators are attached to muscles or tendons, commonly the gastrocnemius or tibialis anterior, to stimulate muscle spindles and create an altered sense of body or joint movement (Fransson et al., 1998; Johansson et al., 2001; Johansson & Magnusson, 1991). The resulting compensatory postural adjustments, captured through centre-of-pressure trajectories, provide insight into how strongly the postural control system relies on proprioceptive cues and how effectively it adapts when these cues are disrupted (Assländer & Peterka, 2014; Fransson et al., 1998; Johansson & Magnusson, 1991). These approaches highlight both the importance of proprioceptive integrity for stable stance and the increased vulnerability of older adults

when somatosensory information becomes degraded (Manchester et al., 1989; Sturnieks et al., 2008).

Visual contributions to balance are often assessed by removing or manipulating visual input. Eyes-closed conditions remove visual information and typically increase postural sway, particularly in older adults or when somatosensory information is also reduced (Manchester et al., 1989; Sturnieks et al., 2008). Perturbed visual environments, such as moving visual surrounds or optic-flow stimuli, further challenge orientation by creating conflict between visual motion cues and body-position information from somatosensory and vestibular systems (Assländer & Peterka, 2014; Fransson et al., 1998; Peterka, 2018). These paradigms are useful because they examine visual dependence and sensory reweighting, rather than simply measuring balance under normal visual conditions (Assländer & Peterka, 2014; Peterka, 2018).

Vestibular contributions can be evaluated using tasks or laboratory methods that increase reliance on head-orientation and gravitational cues. Clinical or functional approaches may include standing or walking while turning or nodding the head, because these tasks require the vestibular system to contribute to gaze stabilisation, head orientation, and postural equilibrium (Agrawal et al., 2009; Horak, 2006; Wagner et al., 2021). Vestibular contribution can also be inferred during balance tasks performed on compliant surfaces or with reduced visual input, because limiting somatosensory or visual information increases reliance on vestibular orientation cues (Horak, 2006; Manchester et al., 1989; Peterka, 2018; Shaffer & Harrison, 2007). More specialised laboratory or clinical methods, including vestibular function testing, galvanic vestibular stimulation, or rotational stimuli, can further characterise vestibular sensitivity and postural responses to vestibular perturbation (Agrawal et al., 2009; Hansson & Magnusson, 2013; Wagner et al., 2021). Age-related vestibular dysfunction and vestibular asymmetry have been associated with dizziness, imbalance, and increased falls-risk in older adults, although vestibular contributions are often difficult to isolate because visual, somatosensory, and vestibular inputs interact during most balance tasks (Agrawal et al., 2009; Hansson & Magnusson, 2013; Wagner et al., 2021).

Overall, sensory-specific assessments reveal how older adults regulate balance when visual, somatosensory, or vestibular information is altered, reduced, or placed in conflict (Assländer & Peterka, 2014; Horak, 2006; Manchester et al., 1989; Peterka, 2018). By examining

postural stability under modified sensory conditions, researchers can identify which sensory inputs are relied upon most strongly, how effectively sensory reweighting occurs when information becomes less reliable, and where potential vulnerabilities in the postural control system may emerge (Assländer & Peterka, 2014; Peterka, 2018; Teasdale et al., 1991).

Collectively, falls-risk screening, clinical testing, instrumented balance assessment, and sensory-specific methods each provide different perspectives on instability in older adults. These approaches highlight the multidimensional nature of falls-risk, but they also raise an important question: whether static balance measures, dynamic gait measures, and falls-risk screening classifications reflect overlapping risk profiles or capture distinct dimensions of falls-related impairment. The subsequent section thus establishes a more direct connection between static and dynamic balance assessment and fall risk, thereby providing the rationale for the present thesis.

2.5 Linking Static and Dynamic Balance to Falls-Risk in Older Adults

The preceding sections show that falls-risk is multifactorial and that balance can be assessed across different postural and movement contexts. A key question arising from this literature is whether static and dynamic balance measures reflect the same underlying impairment or capture complementary dimensions of postural control. This distinction is important because falls often occur during movement-based and transitional activities, while many balance assessments are conducted under controlled standing conditions (Maki, 1997; Robinovitch et al., 2013). Such transitions require adjustments in centre-of-mass position, weight transfer, and the base of support, placing additional demands on postural control during the initiation of walking (Åberg et al., 2010; Buckley et al., 2009). In addition to these postural demands, moving from sitting to standing requires cardiovascular adjustments, with impaired blood-pressure regulation following standing associated with poorer gait performance in older adults (Briggs et al., 2020; Tran et al., 2021). Furthermore, Nakanowatari, Hoshi, Sone, et al. (2024) found that neither 10MWT speed nor TUG completion time differed significantly between older adults with and without a history of falls, whereas peak-force asymmetry measured using in-shoe sensors was significantly greater among those with a falls history. These findings highlight the potential value of examining between-limb loading asymmetry, as it may reveal differences in gait mechanics

associated with falls history that are not reflected in overall walking speed or completion time. Accordingly, comparing gait asymmetry following brief periods of mobility and immobility may help determine whether a period of sitting influences subsequent differences in lower-limb loading and gait timing. More broadly, examining the relationships between static balance and gait asymmetry may help clarify whether poorer postural control during standing is reflected in differences in lower-limb loading and timing during walking, and whether these measures provide complementary information about falls-related vulnerability in older adults.

Evidence directly comparing static and dynamic balance suggests that associations between these measures are generally small or inconsistent. In a systematic review and meta-analysis, Kiss et al. (2018) examined associations between static steady-state, dynamic steady-state, proactive, and reactive balance tasks across the lifespan. In older adults, the pooled association between static and dynamic steady-state balance was small ($r = 0.31$). Similarly, Rizzato et al. (2021) assessed static balance using three 30-second bipedal quiet-standing trials on a force platform, with ellipse area, sway path, and anterior-posterior oscillation used as centre-of-pressure outcomes. Dynamic balance was assessed using an anterior-posterior oscillating platform, in which participants controlled the platform's movement towards visual targets, and performance was quantified using a total stability index. Across these measures, Rizzato et al. (2021) found no significant correlations between static and dynamic balance, with correlations ranging from $r = 0.135$ to $r = 0.202$ during single-task testing, and from $r = -0.099$ to $r = -0.008$ during dual-task testing. Therefore, these findings suggest that quiet-standing and movement-based balance assessments should not be assumed to provide equivalent information, even though they may rely on shared postural control systems.

Furthermore, the limited agreement between static and dynamic balance measures is consistent with contemporary falls-prevention guidance, which supports multidomain assessment rather than reliance on a single screening question, balance test, or mobility measure (Montero-Odasso et al., 2022). However, staged assessment pathways commonly begin with brief screening or case-finding tools, such as the Three Key Questions or STEADI, before progressing to objective gait, balance, or multifactorial assessment when risk is identified (Montero-Odasso et al., 2022). Although this method is practical, subsequent

assessments may depend on whether an individual is initially classified as at risk. This necessitates an investigation into how falls-risk screening classifications correlate with objective balance measurements, as well as whether static and dynamic balance evaluations identify comparable or distinct facets of postural control. The present thesis addresses this gap by comparing STEADI and static balance-derived falls-risk classifications, examining gait asymmetry across mobility and immobility conditions and static balance-derived risk groups, and investigating relationships between static balance measures and dynamic gait asymmetry in community-dwelling older adults. Consequently, Chapter Three presents the research aims and hypotheses developed from this body of literature.

Chapter Three: Research Aims and Hypotheses

The literature review presented in Chapter Two explored falls-risk in older adults and examined the contributions of questionnaire-based screening, static balance assessment, and dynamic gait analysis in evaluating falls-related vulnerability. These approaches provide different perspectives: screening tools identify broad self-reported and clinically recognised falls-risk indicators, static balance tests assess postural stability under controlled conditions, and gait assessments examine loading and temporal symmetry during movement. However, comparatively little research has examined the extent to which falls-risk classifications derived from questionnaire-based screening and objective balance assessment agree, or how these classifications relate to measures of static and dynamic balance. Further investigation is also needed to determine whether gait symmetry differs after periods of mobility and immobility, and whether these responses vary between community-dwelling older adults classified as Low- or High-falls risk using an objective static balance-derived classification. These gaps informed the overarching research question: To what extent do questionnaire-based falls-risk screening, static balance assessment, and dynamic gait analysis provide overlapping or complementary information about falls-related vulnerability in community-dwelling older adults? Accordingly, the following aims and hypotheses were developed for this study.

3.1 Aims

This thesis aimed to:

1. Classify community-dwelling older adults into Low- and High-risk falls groups using the validated Stopping Elderly Accidents, Deaths and Injuries (STEADI) Stay Independent questionnaire, developed by the United States Centres for Disease Control and Prevention.
2. Assess static balance using force-plate-derived centre-of-pressure variables, derive a static balance-based falls-risk classification, and examine its agreement with STEADI screening classification.
3. Assess gait symmetry following periods of mobility and immobility using the absolute symmetry index, and determine whether gait symmetry differs between test condition and between static balance-derived Low- and High-risk groups.
4. Examine the relationships between selected static and dynamic balance measures and determine whether selected dynamic gait-asymmetry variables predict static balance-derived fall-risk classification.

3.2 Hypothesis

1. It was hypothesised that STEADI screening classification would demonstrate limited agreement with the static balance-derived falls-risk classification, with objective static balance assessment identifying some participants with elevated postural sway who were classified as Low-risk by STEADI.
2. It was hypothesised that static and dynamic balance measures would be related but not interchangeable, such that static balance-derived falls-risk classification would be associated with differences in selected dynamic gait-asymmetry variables but would not account for dynamic balance performance across all measured outcomes.
3. It was hypothesised that gait asymmetry would differ between the Mobile-10-Metre Walk Test and Immobile-10-Metre Walk Test conditions and between static balance-derived Low- and High-risk groups, with greater asymmetry expected following the period of immobility and among participants classified as High-risk.

Chapter Four: Methodology

4.1 Research Design

This study employed a mixed, correlational design to investigate the relationships among falls-risk screening, static balance, and dynamic balance in community-dwelling older adults. A within-subject approach was used, whereby each participant completed all components, including a self-reported falls-risk assessment and both static and dynamic balance assessments. The assessment protocols were developed and implemented in collaboration with Sport Manawatū and its Active Living Team to ensure that the procedures were appropriate for the target population and feasible within the community-based testing environment.

4.2 Participants

Thirty-six community-dwelling older adults were recruited through Sport Manawatu's Active Living and Green Prescription programmes. Of these, thirty-three participants were included in the final analysis. Their characteristics, expressed as mean ± SD (range) were: aged 73.94 ± 4.83 (65-82) years; height 163.4 ± 8.27 (145-179) cm; body mass 79.72 ± 16.67 (41-116) kg; and BMI 29.74 ± 5.42 (19.7-40.4) $kg \cdot m^{-2}$. Height and body mass were measured on the day of testing as part of the anthropometric assessment procedures, as discussed in section 4.4.

Participants eligibility was determined using the following inclusion and exclusion criteria.

- Inclusion criteria: Participants were required to be aged 65 years or older, living independently in the community, and able to walk without a walking aid or other assistive device. They were also required to understand the study information, provide informed consent, follow the testing instructions, and complete the static and dynamic balance assessments safely.
- Exclusion criteria: Participants were excluded if they had an acute musculoskeletal injury or a current medical condition that could affect standing or walking performance, compromise their safety, or prevent completion of the testing procedures. Participants were also excluded if they required an assistive device for walking or experienced pain, dizziness, acute illness, or other symptoms on the day of testing that made participation unsafe.

Prior to participation, individuals were provided with a participant information sheet outlining the purpose of the research, the testing procedures, and the potential risks and benefits associated with participation (Appendix One). Participants then completed a written consent form (Appendix Two) and a health screening questionnaire to confirm their eligibility to participate safely in the study (Appendix Three).

Ethical approval for the study was granted by the Massey University Human Ethics Committee: Ohu Matatika 1, Application 25/76 (Appendix Four).

4.3 Priori Statistics

A priori sample size estimation was conducted using G*Power statistical analysis software (version 3.1.9.7; Heinrich Heine University Düsseldorf, Düsseldorf, Germany), informed by previous research examining differences between fallers and non-fallers in older adult populations. For static balance measures, an effect size of 0.23 was reported for postural sway during eyes-closed standing (Johansson et al., 2017), indicating a substantially large sample size requirement (n = 986) to achieve adequate statistical power.

In contrast, previously reported effect sizes for gait asymmetry measures during dynamic walking ranged from 2.37 to 3.95, suggesting that a much smaller sample (n = 10) would be sufficient to detect meaningful differences (Nakanowatari, Hoshi, Sone, et al., 2024). Given this disparity, gait asymmetry variables were prioritised as the primary outcome measures, while static balance measures were included as exploratory comparisons.

Based on these considerations, a target sample size of 30-60 participants was initially proposed to balance statistical power, feasibility, and participant commitment. The final sample size of 36 participants fell within this range and was considered appropriate given the exploratory nature of the study and the large effect sizes associated with the primary dynamic balance outcomes.

4.4 Procedure

Data collection was conducted at Barber Hall Arena in Palmerston North, New Zealand. Upon recruitment, participants were given the option to complete the informed consent form and screening questionnaires online prior to attending their scheduled testing session. For those who had not completed these documents in advance, consent and screening were completed with the researcher upon arrival, prior to the commencement of any testing procedures.

Following confirmation of consent and eligibility, anthropometric measurements were obtained. Body mass was measured using digital scales (Tanita BC-533, Tokyo, Japan), and standing height was recorded using a portable stadiometer (SECA 213, Hamburg, Germany). Participants completed the STEADI Falls-Risk Screening Tool either online before attending their testing session or upon arrival, depending on their preferred method of questionnaire completion. The tool was used to identify recognised falls-risk factors and classify participants according to their level of falls risk (Appendix Five). Developed by the CDC, STEADI supports the early identification of falls risk and the implementation of evidence-based falls-prevention strategies among community-dwelling older adults (CDC, 2023; Lohman et al., 2017). Adapted STEADI screening algorithms have also demonstrated predictive validity for future falls in older adult populations, supporting the use of the tool in the present study (Kannoth et al., 2021; Stevens & Phelan, 2013).

Participants were subsequently fitted with standardised footwear (Figure 3c, H&H Adults' Unisex Freestyle Canvas Low Top Sneakers, The Warehouse Limited, New Zealand) containing wireless Loadsol® Pro-t in-shoe force sensors Figure 3A (Novel GmbH, Munich, Germany). Each shoe housed an insole sensor, with the battery unit secured to the upper portion of the shoe via the laces (Figure 3B). The insoles were worn throughout all testing procedures; however, force data were only recorded during dynamic tasks.



Figure 3. *Loadsol Pro-t insoles and Standardised footwear. A. The LoadSol® Pro-t insole, showing anterior, central, and posterior sections; B. Standardised shoe fitted with the Loadsol® insole, with the battery box attached to the laces (top view); C. Standardised shoe fitted with the Loadsol® insole, with the battery box attached to the laces (side view) (Loadsol® - Mobile force measurement for real-world applications, 2025).*

Prior to data collection, the Loadsol insoles were calibrated according to the manufacturer's recommended protocol using a bipedal stance to ensure accurate distribution of body weight across both sensors. Participant body mass was first entered into the mobile phone application (loadapp, novel GmbH, Munich, Germany). For calibration, the participant then unloaded the left leg while the researcher zeroed the left insole, and the same procedure was repeated for the right leg. The Loadsol system communicated via Bluetooth with the mobile application, which recorded vertical force-time data continuously from both limbs throughout the dynamic balance tasks at a set sample rate of 200 Hz. Continuous force recording across multiple steps enabled the extraction of spatiotemporal variables (stride duration, stride frequency, ground contact time, swing time, peak impact time, midstance time, and time to active peak force) and kinetic variables (peak impact force, average loading rate, midstance force, active peak force, and impulse), allowing the identification of gait characteristics and between-limb asymmetries during walking. Raw data were subsequently exported and processed using custom MATLAB scripts (R2024a, MathWorks, Inc., Natick, MA, USA).

Following device setup, participants completed the assessments in a standardised order, beginning with the static balance assessment using the BTrackS Balance Plate (Balance Tracking Systems Inc., San Diego, CA, USA). This was followed by the Mobile-10MWT and then the Immobile-10MWT, with both dynamic balance tasks assessed using Loadsol in-shoe force sensors. All participants completed the assessments according to the same standardised protocols.

Each testing session lasted approximately 15 minutes per participant. Throughout all procedures, participants were closely monitored, and a member of the research team remained nearby to ensure safety and provide assistance if required.

4.5 Static Balance Assessment

Participants stood on the force plate wearing the standardised footwear, with their feet positioned approximately hip-width apart and their hands resting on their hips. They were instructed to face forward and keep their eyes closed throughout the assessment to minimise visual input and increase reliance on the somatosensory and vestibular systems (Baloh et al., 1993; Cheung & Schmuckler, 2021; Peterka, 2018). BTrackS Balance Plate and example software outputs are shown in Figure 4. Each participant completed one 20-second familiarisation trial, followed by three 20-second trials used for analysis. Participants remained on the force plate between trials.

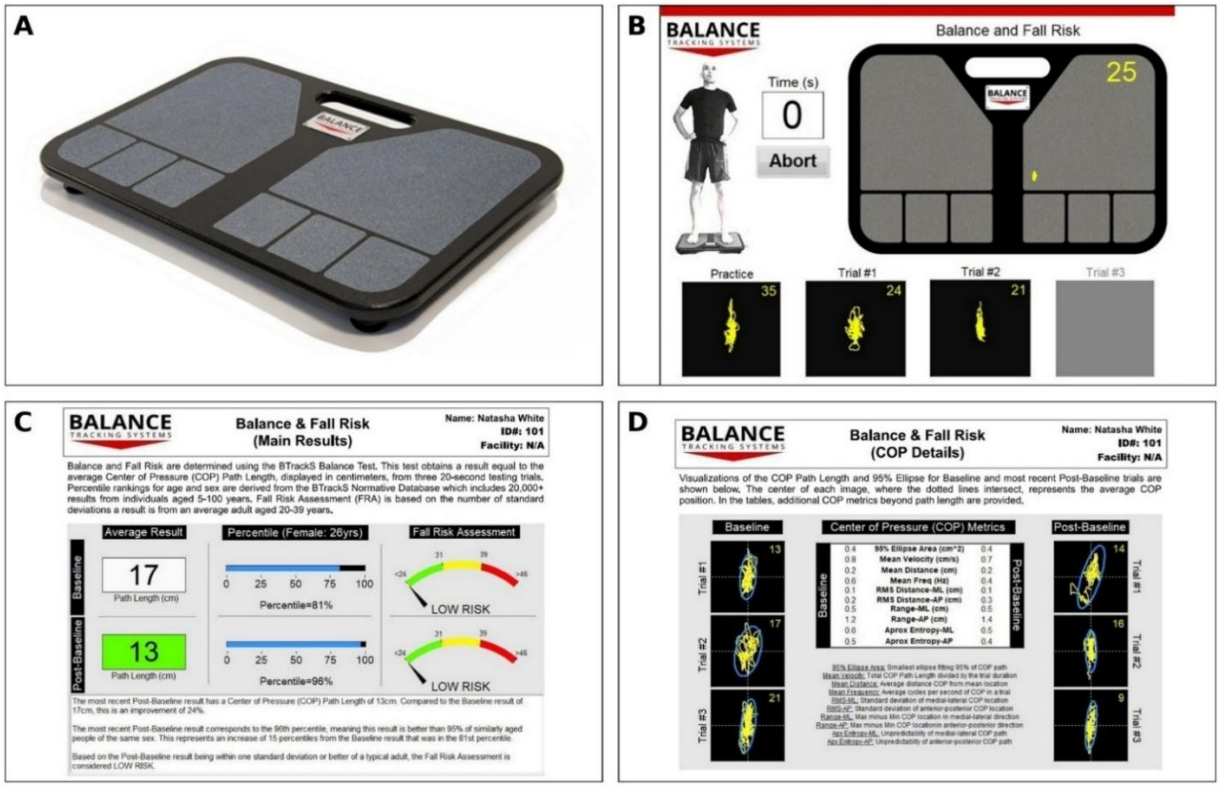


Figure 4. *BTrackS static balance assessment equipment and representative software outputs. A. BTrackS Balance Plate; B. BTrackS Balance software interface displaying trial collection; C. Example main results output, including centre-of-pressure path length and fall-risk assessment; D. Example centre-of-pressure details output showing trial plots and additional centre-of-pressure variables*

The BTrackS Balance software automatically calculated an average across the three recorded trials to provide a representative measure of postural stability. While the system generated a range of sway-related variables, including those displayed in Figure 4 C-D, the primary outcome of interest in the present study was centre-of-pressure path length. This variable represents the total distance travelled by the centre-of-pressure during quiet standing and is widely used as an indicator of postural sway magnitude (Goble et al., 2019; O'Connor et al., 2016).

4.6 Dynamic Balance Assessment

Two functional mobility tasks were used to assess dynamic balance: a Mobile 10-Metre Walk Test (Mobile-10MWT) and an Immobile 10-Metre Walk Test (Immobile-10MWT). Both tasks were performed on the wooden floor of Barber Hall while participants wore standardised footwear fitted with Loadsol in-shoe force sensors.

4.6.1 Mobile-10MWT

The Mobile-10MWT was conducted, with cones marking the start and finish of the 10-metre walkway. Prior to data collection, participants completed 1-minute of walking at their pace of choice as a warm-up. Participants then stood behind the start marker for a very brief period (<10s) and were instructed to walk towards the finish marker at their normal walking pace. Plantar force data logging commenced when the participant crossed the start marker and stopped once they had passed the finish marker (Figure 5A).

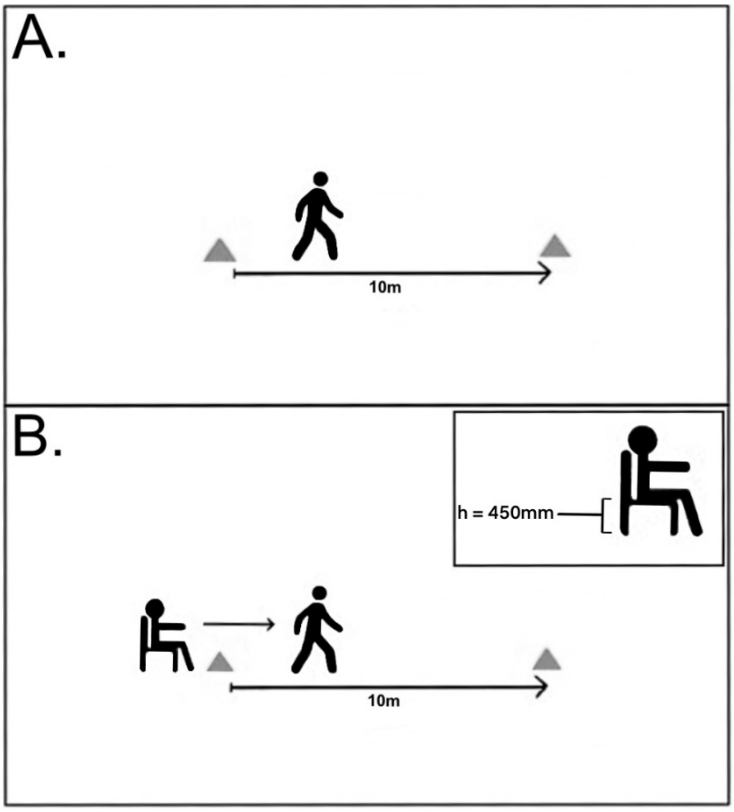


Figure 5. *Dynamic balance task set-ups. A. Mobile-10MWT and B. Immobile-10MWT, with chair height represented by h*

4.6.2 Immobile-10MWT

The Immobile-10MWT was conducted on a parallel pathway adjacent to the Mobile-10MWT to allow efficient transition between tasks. The set-up mirrored the walking test, with cones marking the start and finish of the 10-metre pathway and a chair (height = 450 mm) positioned directly behind the starting marker (Figure 5B).

Participants began seated on the chair and remained seated for one minute prior to initiating the task. Following this period, they were instructed to stand and walk towards the finish marker at their normal walking pace. Data logging commenced on the first step after the participant rose from the chair and crossed the start marker. The same gait parameters derived from the Mobile-10MWT were extracted from the Immobile-10MWT and used to assess symmetry between the left and right legs.

4.7 Data Processing

4.7.1 Static Balance Data Processing

All dependent variables for static balance were processed using the BTrackS Assess Balance software (Balance Tracking Systems, Inc., San Diego, CA, USA), a Windows-based application that interfaces with the BTrackS Balance Plate and generates standardised balance reports. As shown in the exported BTrackS report (Figure 4), centre-of-pressure path length was reported as the average across three 20-second trials alongside percentile rankings and fall-risk classifications. The BTrackS fall-risk classifications were recorded for each participant and later compared with STEADI fall-risk outcomes.

4.7.2 Dynamic Balance Data Processing

Dynamic balance force-time data were pre-processed according to a standardised protocol. A 5-point moving-average filter was then applied to smooth the force-time signal and reduce high-frequency variability. All processed signals were visually inspected to ensure quality and consistency across trials, and the start and end points of each trial were manually selected to retain steady-state walking data. Finally, the data were resampled to 1000 Hz to improve temporal resolution and enable more precise identification of gait events (Chan et al., 2018; Macdermid et al., 2025). Initial contact and toe-off events were identified using force thresholds of 10-20 N, allowing segmentation of individual steps within each trial (Macdermid et al., 2025; Wright et al., 1998). From these identified gait events, both spatiotemporal and kinetic variables were calculated.

Kinetic variables derived from the vertical force-time curve, together with the gait events used to identify time-based gait variables, are illustrated in Figure 6.

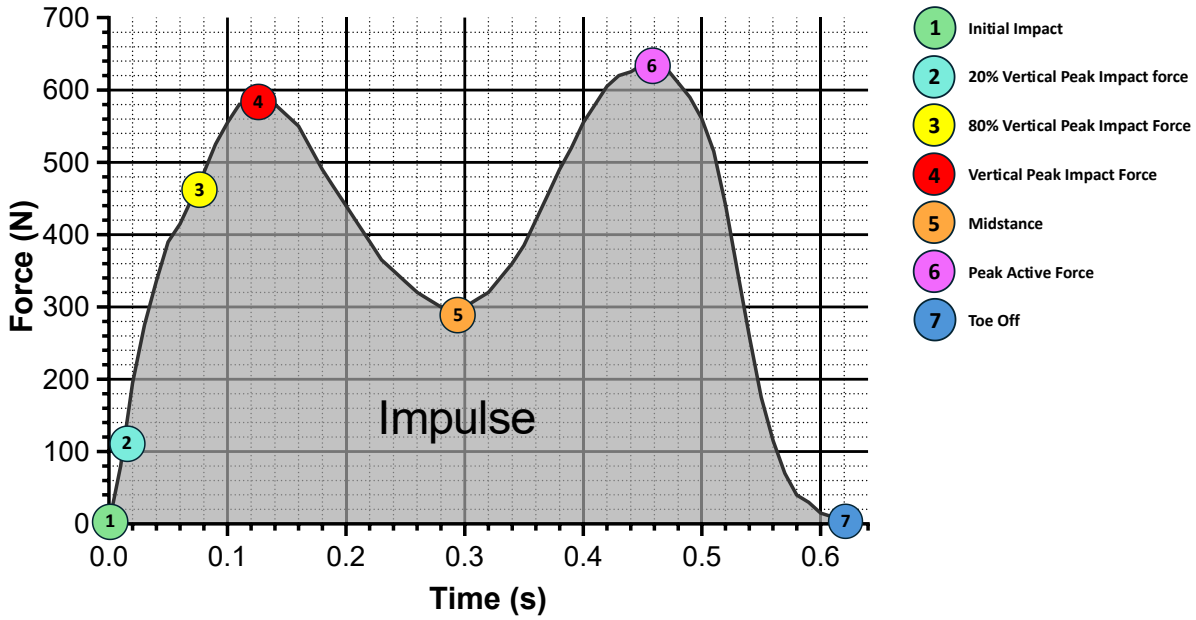


Figure 6. *Representative vertical force-time curve for one step, showing the kinetic variables and gait events identified from the total vertical force signal during walking*

Spatiotemporal variables described the temporal characteristics of walking based on initial contact and toe-off events. Ground contact time (s) was defined as the interval from initial contact to toe-off of the same foot, representing the period of foot-ground contact during each step. Swing time (s) was defined as the interval from toe-off to the next initial contact of the same foot, representing the period during which the foot was not in contact with the ground. Stride duration (s) was defined as the time between two successive initial contacts of the same foot, representing one complete gait cycle. Stride frequency (strides·min^{-1}) was calculated as the number of strides completed per minute, reflecting cadence.

Kinetic variables described the vertical forces generated during walking, as shown in Figure 6. Peak impact force (N·BW^{-1}) was defined as the first force peak occurring shortly after initial contact, representing the initial loading response following foot-ground contact. Average loading rate (N·BW^{-1}·s^{-1}), described the rate at which force increased following initial contact and was calculated as the average slope of the force-time curve between 20% and 80% of the first impact peak. Peak active force (N·BW^{-1}) was defined as the second force peak occurring during late stance, representing the propulsive phase of gait associated with push-off. Time to peak active force (s) was defined as the interval from initial contact to peak active force, reflecting the timing of the second force peak during late stance. Midstance

force ($N \cdot BW^{-1}$) corresponded to the trough between peak impact force and peak active force, reflecting the relative unloading period as the body progressed over the supporting foot. Midstance force ($N \cdot BW^{-1}$) was defined as the trough between peak impact force and peak active force during stance, while midstance time (s) represented the time from initial contact to this trough. Impulse ($N \cdot s \cdot BW^{-1}$) was calculated as the area under the force-time curve during the stance phase, representing the cumulative vertical load applied during each step. All force-related variables were normalised to body weight to account for inter-individual differences and enable comparison across participants.

4.7.2 Symmetry Index Calculation

Dynamic balance was quantified using a symmetry index (SI), calculated as the difference between the left and right legs for each spatiotemporal and kinetic variable. The symmetry index was expressed as a percentage using Equation 1.

Equation 1. Symmetry index calculation

$$SI\% = \left[\frac{(L - R)}{0.5(L + R)}\right] \cdot 100$$

Positive SI values indicated a greater contribution from the left leg, and negative SI values indicated a greater contribution from the right leg. Values closer to zero reflected a more symmetrical distribution between legs, while values further from zero indicated greater between-leg asymmetry.

Separate SI datasets were generated for both the Mobile-10MWT and Immobile-10MWT and were kept for individual-level reporting, as they preserved the direction of asymmetry.

However, because the direction of asymmetry varied across participants, absolute symmetry index (ASI) values were calculated prior to group-level statistical analysis to remove directional bias associated with left or right leg dominance. The ASI was also expressed as a percentage and calculated using Equation 2.

Equation 2. Absolute symmetry index calculation

$$ASI\% = \left|\left[\frac{L - R}{0.5(L + R)}\right]\right| \cdot 100$$

This transformation allowed asymmetry to be expressed solely in magnitude, such that larger ASI values indicated greater between-leg differences regardless of direction.

4.7.3 STEADI Fall-Risk Classification

Participants completed the STEADI falls-risk screening questionnaire (Appendix Four), which classified individuals based on reported falls history, perceived unsteadiness, concern about falling, and other recognised falls-risk factors. Scores were calculated according to standard guidelines, with participants ordinarily classified as not at risk (score 0-3) or at risk (score ≥ 4). For the purposes of the present study and to provide consistent terminology for risk-based group comparisons, these classifications were relabelled as Low-risk and High-risk, respectively. STEADI classifications were recorded for each participant and used to describe participants and compare them with the falls-risk classification derived from the static balance assessment.

4.8 Statistical analysis

All statistical analyses were performed using GraphPad Prism (version 10.6.1; GraphPad Software, San Diego, CA, USA), except for the binary logistic regression, which was conducted using IBM SPSS Statistics (version 23, IBM, New York, USA). Statistical significance was set at $p < 0.05$ for all analyses. Participant characteristics were summarised using descriptive statistics and reported as mean ± standard deviation (SD) and range, while static and dynamic balance variables were reported as mean ± SD.

Static balance variables obtained from the BTrackS Assess Balance software were first summarised descriptively for the total sample. The proprietary BTrackS fall-risk classification was then recorded for each participant. The software classified participants into Low-, Moderate-, and High-risk categories, the Moderate- and High-risk categories were combined to create a binary static balance-derived classification for comparison with the binary STEADI classification and for subsequent group-based analyses. Directional disagreement between STEADI and static balance-derived falls-risk classifications was examined using McNemar's test, with the exact McNemar test also reported. Cohen's kappa was used to assess the level of agreement between methods.

Participants were subsequently grouped according to the static balance-derived falls-risk classification for between-group analyses of static balance variables. Prior to inferential testing, variables were assessed for normality using the D'Agostino-Pearson normality test. Between-group differences in static balance variables were examined using independent-samples *t*-tests when data were normally distributed and Mann-Whitney U tests when the normality assumption was not met.

Dynamic balance was quantified using absolute symmetry index (ASI%) values for each spatiotemporal and kinetic variable under both the Mobile-10MWT and Immobile-10MWT conditions. To examine the effects of test condition and falls-risk group on dynamic asymmetry, two-way mixed-effects analyses were performed for each ASI variable, with test condition (Mobile-10MWT vs Immobile-10MWT) as the within-subject factor and static balance-derived falls-risk group (Low-risk vs High-risk) as the between-subject factor. Main effects of test condition and falls-risk group, together with the test condition*falls-risk group interaction, were examined. Where relevant, post-hoc pairwise comparisons adjusted for multiple comparisons were used to further explore group differences within each condition. Corresponding left- and right-foot values were also reported descriptively for each test condition and fall-risk group.

To examine whether selected dynamic asymmetry variables predicted static balance-derived falls-risk classification, a binary logistic regression was performed in SPSS. Falls-risk classification was coded as Low-risk = 0 and High-risk = 1. Predictor variables included mean swing time ASI%, and mean average loading rate ASI%. These variables were selected because they differed between falls-risk groups, had complete data, and were considered sufficiently independent for inclusion in the model. Mean values across the Mobile-10MWT and Immobile-10MWT conditions were used because no main effect of test condition was identified for either variable in the preceding two-way analyses. Overall model fit was assessed using the omnibus test of model coefficients, while model fit was examined using the Hosmer-Lemeshow goodness-of-fit test. The fitted regression equation and overall classification accuracy, including the percentages of correctly classified Low- and High-risk participants, were also reported.

Chapter Five: Results

5.1 STEADI Screening Classification

Thirty-three participants met the inclusion criteria and were included in the final analysis. Based on the STEADI screening, 27 of the 33 participants were classified as Low-falls-risk and 6 as High-falls-risk.

5.2 Static Balance and Falls-Risk

Centre-of-pressure measures obtained during quiet standing with eyes closed showed a mean sway path length of 38.73 ± 12.90 cm, with 95% of the path length samples contained within an ellipse area of 2.93 ± 2.78 cm^2. Mean path length velocity was 1.94 ± 0.65 $cm \cdot s^{-1}$, with a sway frequency of 0.55 ± 0.14 Hz. The average displacement from the mean position was 0.60 ± 0.21 cm. Directional sway magnitude quantified using root-mean-square (RMS) displacement was 0.67 ± 0.22 cm in the anterior-posterior direction and 0.22 ± 0.13 cm in the medial-lateral direction. Total sway range was 3.32 ± 1.18 cm in the anterior-posterior direction and 1.14 ± 0.56 cm in the medial-lateral direction. Approximate entropy values were 0.47 ± 0.10 in the anterior-posterior direction and 0.51 ± 0.11 in the medial-lateral direction, respectively.

Using the variables presented, the proprietary software calculated falls-risk and classified participants as Low-Moderate-High. There were 12 Low-risk, 3 Moderate-risk and 18 High-risk participants. To enable comparison between STEADI classifications and static balance outcomes, participants were categorised as either Low- or High-falls-risk. McNemar's test indicated a significant directional disagreement between the two classification methods ($\chi^2(1) = 11.53$, $p = 0.0007$), a result supported by the exact McNemar test ($p = 0.0003$). The static balance test classified substantially more participants as High-risk than STEADI, demonstrating a systematic imbalance in how the two methods assign risk. Agreement between the two methods was low, with Cohen's kappa (k) = 0.12, indicating only slight agreement.

Separating static balance test data into the respective groups highlighted significant differences between groups in centre-of-pressure-related variables. The normality of each variable was assessed before between-group comparisons were performed. Sway path

length was normally distributed in both the Low-risk (K^2 = 2.69, p = 0.260) and High-risk (K^2 = 0.97, p = 0.617) groups and differed between groups ($t_{(31)}$= 8.03, p < 0.0001, Figure 7A), with a path length of 24.92 ± 3.78 cm for the Low-risk group and a path length of 46.62 ± 8.86 cm for the High-risk group. Normal distributions were also observed for sway path length velocity in both the Low-risk (K^2 = 0.93, p = 0.630) and High-risk (K^2 = 0.81, p = 0.668) groups. Velocity differed between groups ($t_{(31)}$= 8.28, p < 0.0001), with values of 1.23 ± 0.19 cm·s^{-1} for the Low-risk group and 2.34 ± 0.44 cm·s^{-1} for the High-risk group (Figure 7B). Likewise, frequency was normally distributed in both the Low-risk (K^2 = 0.52, p = 0.769) and High-risk (K^2 = 3.51, p = 0.173) groups and differed ($t_{(31)}$= 3.37, p = 0.002), with a frequency of 0.45 ± 0.13 Hz for the Low-risk group and a frequency of 0.60 ± 0.12 Hz for the High-risk group (Figure 7C).

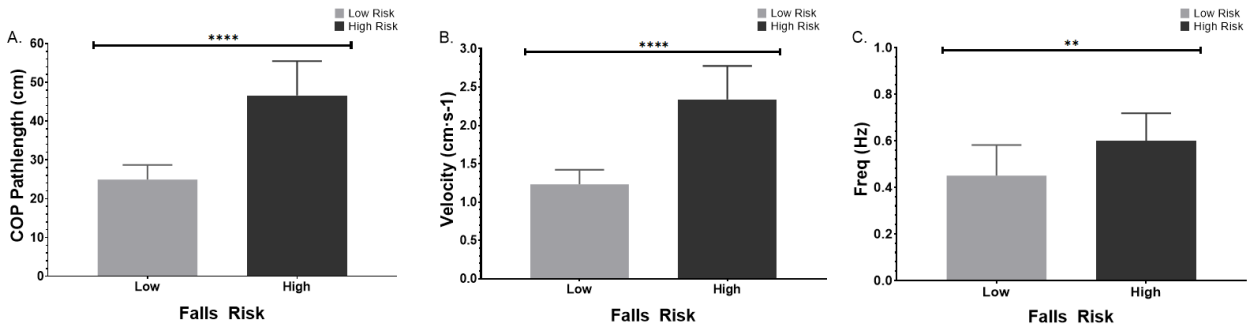


Figure 7*. Mean ± SD of static balance variables by falls-risk group. A. Centre-of-pressure path length (cm); B. Velocity (cm·s^{-1}); C. Frequency (Hz). ** p = < 0.01; **** p = < 0.0001 indicates level of significance*

The predictability of anterior-posterior sway quantified using approximate entropy was normally distributed in both the Low-risk (K^2 = 0.26, p = 0.879) and High-risk (K^2 = 2.12, p = 0.347) groups. Values differed between groups ($t_{(31)}$ = 3.13, p = 0.004), with an entropy of 0.41 ± 0.11 for the Low-risk group and an entropy of 0.51 ± 0.08 for the High-risk group (Figure 8A). In the medial-lateral direction, approximate entropy was normally distributed in both the Low-risk (K^2 = 3.98, p = 0.137) and High-risk (K^2 = 1.61, p = 0.446) groups. However, no difference was observed between groups ($t_{(31)}$ = 0.99, p = 0.329), with values of 0.48 ± 0.11 for the Low-risk group and 0.52 ± 0.11 for the High-risk group. For anterior-posterior root-mean-square displacement, normality was supported in both the Low-risk (K^2 = 5.20, p = 0.074) and High-risk (K^2 = 3.50, p = 0.174) groups and differed between groups ($t_{(31)}$ = 2.54,

p = 0.016), with values of 0.55 ± 0.18 cm for the Low-risk group and 0.74 ± 0.22 cm for the High-risk group (Figure 8B). In contrast, medial-lateral root-mean-square displacement was normally distributed in the Low-risk group (K^2 = 1.40, p = 0.496), whereas the High-risk group was not (K^2 = 29.31, p < 0.0001). Medial-lateral root-mean-square displacement differed between groups (*U* = 67.5, *p* = 0.017), with values of 0.16 ± 0.07 cm for the Low-risk group and 0.25 ± 0.15 cm for the High-risk group (Figure 8C). A similar pattern was found for the 95% ellipse area, which was normally distributed in the Low-risk group (K^2 = 3.81, p = 0.149), but not in the High-risk group (K^2 = 39.34, p < 0.0001). Ellipse area differed between groups (*U* = 59, *p* = 0.011), with values of 1.76 ± 1.04 cm^2 for the Low-risk group and 3.60 ± 3.24 cm^2 for the High-risk group (Figure 8D).

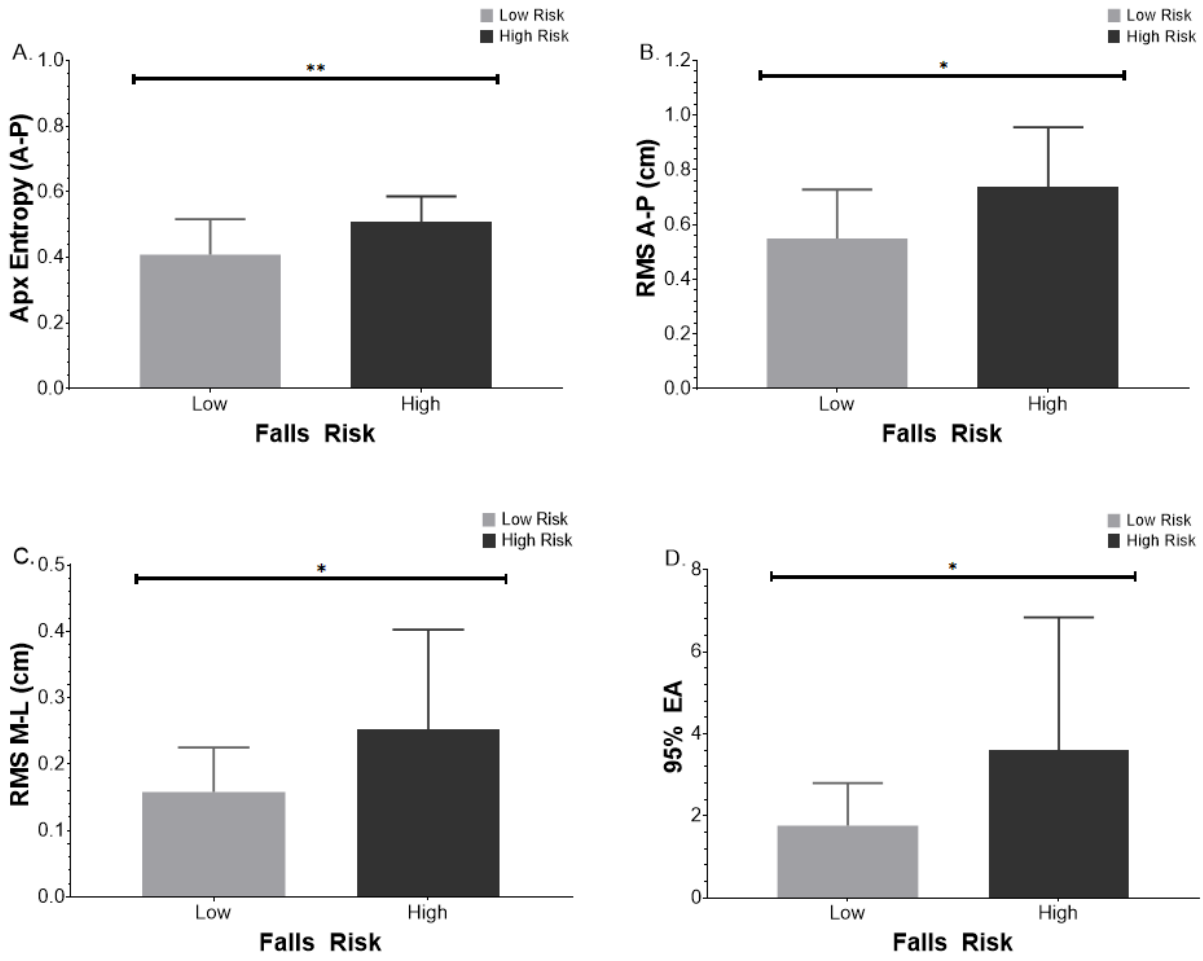


Figure 8. *Mean ± SD of static balance variables by falls-risk group. A. Approximate entropy in the anterior-posterior direction; B. Root-mean-square displacement in the anterior-posterior direction (cm); C. Root-mean-square displacement in the medial-lateral direction (cm); D. 95% ellipse area (cm²). * p = < 0.05; ** p = < 0.01 indicates level of significance*

Anterior-posterior sway range was not normally distributed in the Low-risk group ($K^2 = 6.54$, $p = 0.038$), although it was normally distributed in the High-risk group ($K^2 = 3.45$, $p = 0.179$). Anterior-posterior sway range differed between groups ($U = 53.5$, $p = 0.005$), with an average range of 2.67 ± 0.94 cm for the Low-risk group and 3.69 ± 1.15 cm for the High-risk group (Figure 9A). Similarly, medial-lateral sway range was normally distributed in the Low-risk group ($K^2 = 2.25$, $p = 0.324$), but not in the High-risk group ($K^2 = 24.52$, $p < 0.0001$), and differed between groups ($U = 59.5$, $p = 0.011$), with an average range of 0.87 ± 0.25 cm for the Low-risk group and 1.30 ± 0.63 cm for the High-risk group (Figure 9B). Mean centre-of-pressure distance from the average position followed the same pattern, with a normal distribution in the Low-risk group ($K^2 = 5.20$, $p = 0.074$), but not in the High-risk group ($K^2 = 13.58$, $p = 0.001$). Mean centre-of-pressure distance differed between groups ($U = 52$, $p = 0.004$), with an average distance of 0.48 ± 0.17 cm for the Low-risk group and 0.67 ± 0.21 cm for the High-risk group (Figure 9C).

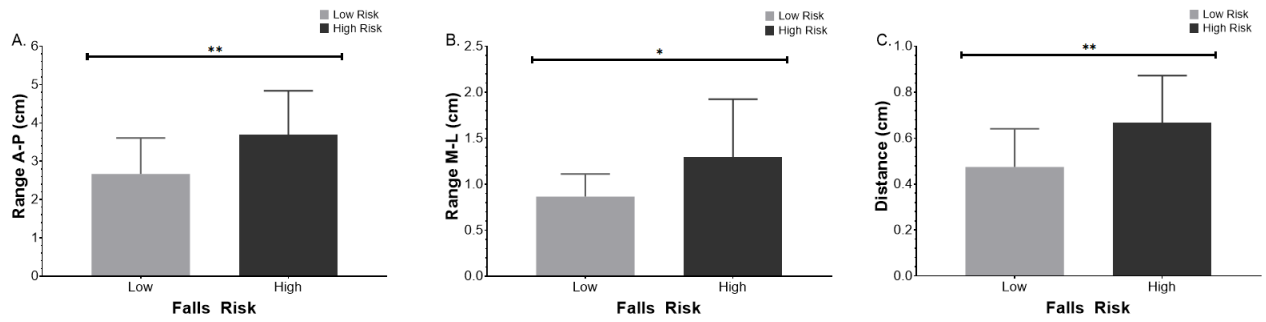


Figure 9. *Mean ± SD of static balance variables by falls-risk group. A. Anterior-posterior range (cm); B. Medial-lateral range (cm); C. Distance (cm). * p = < 0.05; ** p = < 0.01 indicates level of significance*

5.3 Dynamic Balance

5.3.1 Spatiotemporal Variables

Stride duration ASI% for the Mobile-10MWT and Immobile-10MWT in the Low-risk and High-risk groups were 0.68 ± 0.62 % vs. 0.96 ± 1.25 %; and 0.87 ± 0.48 % vs. 0.70 ± 0.59 %, respectively. There was no significant test*falls-risk group interaction for stride duration ($F_{(1, 31)}$ = 1.06, p = 0.310) and no main effects of test ($F_{(1, 31)}$ = 0.03, p = 0.864) or falls-risk group ($F_{(1, 31)}$ = 0.07, p = 0.794). Left and right foot stride duration data for each test are shown in Table 1.

Table 1. *Mean ± SD of stride duration (s) for the left and right foot during the Mobile-10MWT and Immobile-10MWT*

Test	Low-Risk		High-Risk	
	Left	Right	Left	Right
Mobile-10MWT	1.15 ± 0.12	1.15 ± 0.12	1.12 ± 0.13	1.13 ± 0.12
Immobile-10MWT	1.10 ± 0.09	1.10 ± 0.10	1.12 ± 0.11	1.12 ± 0.11

Stride frequency ASI% for the Mobile-10MWT and Immobile-10MWT in the Low-risk and High-risk groups were 0.63 ± 0.94 % vs. 1.06 ± 1.71 %; and 1.04 ± 0.92 % vs. 0.80 ± 0.95 %, respectively. There was no significant test*falls-risk group interaction for stride frequency ($F_{(1, 31)}$ = 1.29, p = 0.264) and no main effects of test ($F_{(1, 31)}$ = 0.06, p = 0.805) or falls-risk group ($F_{(1, 31)}$ = 0.08, p = 0.777). Left and right foot stride frequency data for each test are shown in Table 2.

Table 2. *Mean ± SD of stride frequency (spm) for the left and right foot during the Mobile-10MWT and Immobile-10MWT*

Test	Low-Risk		High-Risk	
	Left	Right	Left	Right
Mobile-10MWT	52.92 ± 5.13	52.92 ± 5.35	54.19 ± 6.03	53.90 ± 5.83
Immobile-10MWT	54.92 ± 4.60	54.83 ± 4.88	54.48 ± 5.16	54.33 ± 5.22

Ground contact time ASI% for the Mobile-10MWT and Immobile-10MWT in the Low-risk and High-risk groups were 2.16 ± 1.30 % vs. 4.11 ± 3.78 %; and 1.85 ± 1.50 % vs. 3.26 ± 2.39 %, respectively (Figure 10). There was no significant test*falls-risk group interaction for ground contact time ($F_{(1, 31)}$ = 0.29, p = 0.595) and no main effects of test ($F_{(1, 31)}$ = 1.31, p = 0.261). However, there was a significant main effect of falls-risk group ($F_{(1, 31)}$ = 4.15, p = 0.050) but no post-hoc differences between Mobile-10MWT Low-risk vs. High-risk (-1.95 ± 0.97, p = 0.09) or Immobile-10MWT Low-risk vs. High-risk (-1.41 ± 0.97, p = 0.28).

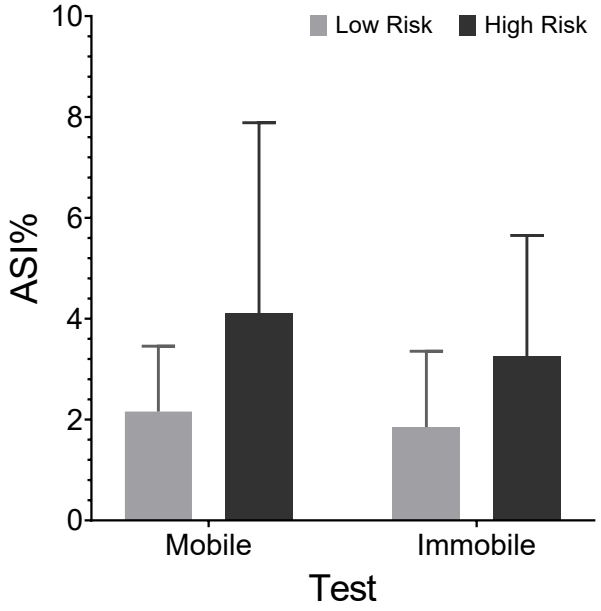


Figure 10. *Mean ± SD of ground contact time ASI% for the left and right foot during the Mobile-10MWT and Immobile-10MWT*

Although ground contact time ASI% differed by falls-risk group overall, this effect was not specific to either test condition. The corresponding left and right foot ground contact time (s) values for each test are presented in Table 3.

Table 3. *Mean* ± SD of ground contact time (s) for the left and right foot during the Mobile-10MWT and Immobile-10MWT

Test	Low-Risk		High-Risk	
	Left	Right	Left	Right
Mobile-10MWT	0.77 ± 0.09	0.78 ± 0.10	0.79 ± 0.10	0.79 ± 0.10
Immobile-10MWT	0.74 ± 0.06	0.75 ± 0.07	0.78 ± 0.10	0.77 ± 0.10

Swing time ASI% for the Mobile-10MWT and Immobile-10MWT in the Low-risk and High-risk groups were 3.87 ± 2.78 % vs. 10.43 ± 12.39 %; and 3.24 ± 2.34 % vs. 8.35 ± 6.99 %, respectively (Figure 11). There was no significant test*falls-risk group interaction ($F_{(1, 31)}$ = 0.24, p = 0.625) and no main effect of test ($F_{(1, 31)}$ = 0.84, p = 0.366). However, there was a significant main effect of falls-risk group ($F_{(1, 31)}$ = 5.11, p = 0.031) but no post-hoc differences between Mobile-10MWT Low-risk vs. High-risk (-6.57 ± 2.98, p = 0.061) or Immobile-10MWT Low-risk vs. High-risk (-5.11 ± 2.98, p = 0.174).

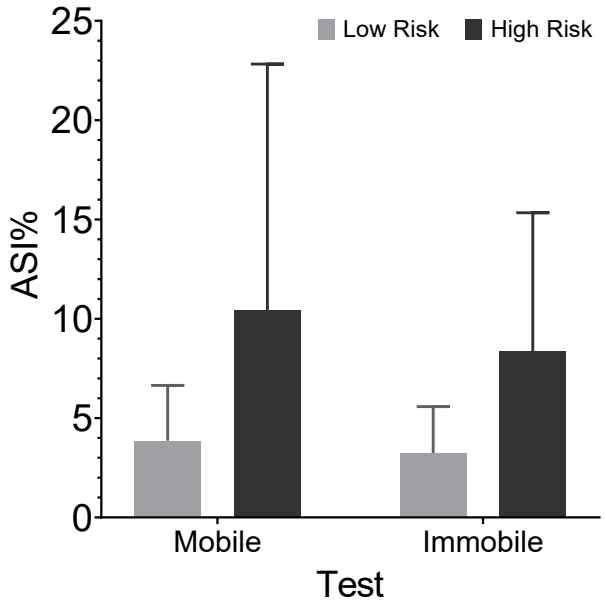


Figure 11. *Mean ± SD of swing time ASI% for the left and right foot during the Mobile-10MWT and Immobile-10MWT*

Although swing time ASI% differed by falls-risk group overall, this effect was not specific to either test condition. The corresponding left and right foot swing time (s) values for each test are presented in Table 4.

Table 4. *Mean ± SD of swing time (s) for the left and right foot during the Mobile-10MWT and Immobile-10MWT*

Test	Low-Risk		High-Risk	
	Left	Right	Left	Right
Mobile-10MWT	0.38 ± 0.04	0.37 ± 0.04	0.34 ± 0.06	0.35 ± 0.06
Immobile-10MWT	0.37 ± 0.03	0.36 ± 0.03	0.34 ± 0.05	0.35 ± 0.05

Peak impact time ASI% for the Mobile-10MWT and Immobile-10MWT in the Low-risk and High-risk groups were 4.68 ± 3.24 % vs. 12.24 ± 8.54 %; and 5.82 ± 3.40 % vs. 13.04 ± 12.38 %, respectively (Figure 12). There was no significant test*falls-risk group interaction ($F_{(1, 31)}$ = 0.01, p = 0.935) and no main effect of test ($F_{(1, 31)}$ = 0.22, p = 0.642). However, there was a significant main effect of falls-risk group ($F_{(1, 31)}$ = 9.46, p = 0.004). Post-hoc comparisons showed greater peak impact time asymmetry in the High-risk group during the Mobile-10MWT (-7.57 ± 3.17, p = 0.040), whereas the corresponding comparison during the Immobile-10MWT did not reach statistical significance (-7.23 ± 3.17, p = 0.052).

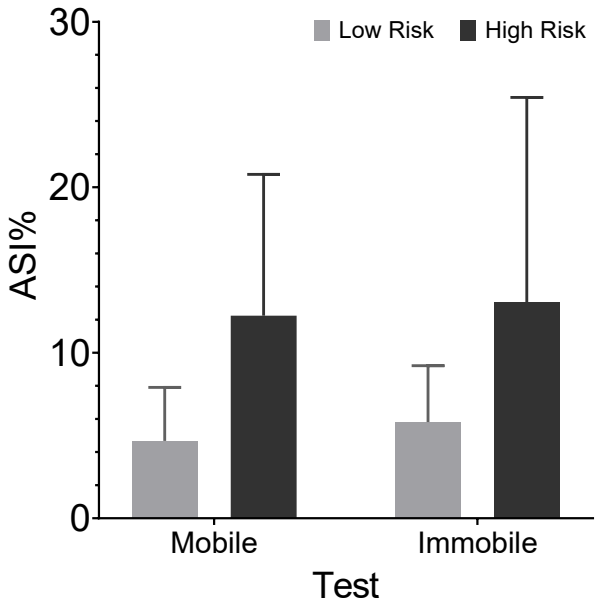


Figure 12. *Mean ± SD of peak impact time ASI% for the left and right foot during the Mobile-10MWT and Immobile-10MWT*

Although peak impact time ASI% differed by falls-risk group overall, this effect was not specific to either test condition. The corresponding left and right foot peak impact time (s) values for each test are presented in Table 5.

Table 5. Mean ± SD of peak impact time (s) for the left and right foot during the Mobile-10MWT and Immobile-10MWT

Test	Low-Risk		High-Risk	
	Left	**Right**	**Left**	**Right**
Mobile-10MWT	0.20 ± 0.05	0.21 ± 0.05	0.21 ± 0.06	0.22 ± 0.08
Immobile-10MWT	0.19 ± 0.04	0.19 ± 0.03	0.21 ± 0.06	0.21 ± 0.06

Midstance time ASI% for the Mobile-10MWT and Immobile-10MWT in the Low-risk and High-risk groups were 4.68 ± 5.37 % vs. 9.03 ± 6.78 %; and 5.42 ± 6.30 % vs. 9.65 ± 8.17 %, respectively (Figure 13). There was no significant test*falls-risk group interaction ($F_{(1, 31)}$ = 0.00, p = 0.972) and no main effect of test ($F_{(1, 31)}$ = 0.17, p = 0.687). However, there was a significant main effect of falls-risk group ($F_{(1, 31)}$ = 5.11, p = 0.031) but no post-hoc differences between Mobile-10MWT Low-risk vs. High-risk (-4.35 ± 2.52, p = 0.171) or Immobile-10MWTLow-risk vs. High-risk (-4.23 ± 2.52 , p = 0.187).

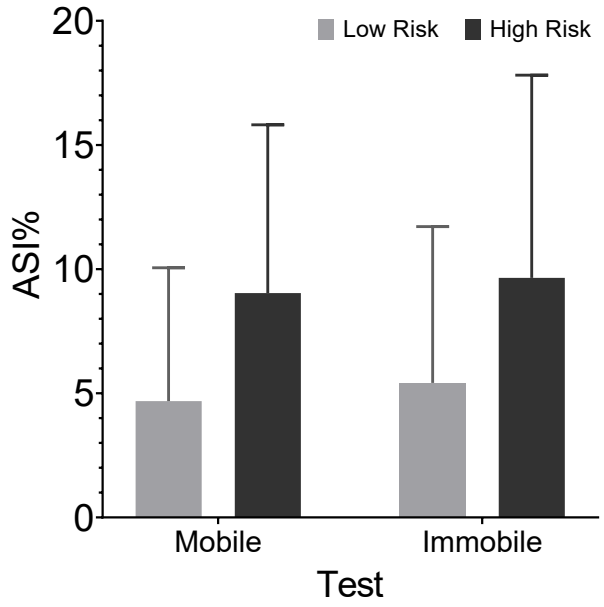


Figure 13. *Mean ± SD of midstance time ASI% for the left and right foot during the Mobile-10MWT and Immobile-10MWT*

Although midstance time ASI% differed by falls-risk group overall, this effect was not specific to either test condition. The corresponding left and right foot midstance time (s) values for each test are presented in Table 6.

Table 6. *Mean ± SD of midstance time (s) for the left and right foot during the Mobile-10MWT and Immobile-10MWT*

Test	Low-Risk		High-Risk	
	Left	Right	Left	Right
Mobile-10MWT	0.33 ± 0.05	0.34 ± 0.04	0.33 ± 0.05	0.35 ± 0.06
Immobile-10MWT	0.33 ± 0.05	0.33 ± 0.03	0.33 ± 0.04	0.34 ± 0.05

Time to active peak force ASI% for the Mobile-10MWT and Immobile-10MWT in the Low-risk and High-risk groups were 2.86 ± 1.86 % vs. 5.98 ± 5.50 %; and 3.79 ± 5.99 % vs. 6.21 ± 6.11%, respectively. There was no significant test*falls-risk group interaction for time to active peak force ($F_{(1, 31)}$ = 0.17, p = 0.686) and no main effects of test ($F_{(1, 31)}$ = 0.46, p = 0.502) or falls-risk group ($F_{(1, 31)}$ = 2.54, p = 0.121). Left and right foot time to active peak force (s) data for each test are shown in Table 7.

Table 7. *Mean ± SD of time to active peak force (s) for the left and right foot during the Mobile-10MWT and Immobile-10MWT*

Test	Low-Risk		High-Risk	
	Left	Right	Left	Right
Mobile-10MWT	0.57 ± 0.04	0.58 ± 0.05	0.55 ± 0.07	0.57 ± 0.06
Immobile-10MWT	0.54 ± 0.05	0.55 ± 0.04	0.55 ± 0.07	0.56 ± 0.06

5.3.2 Kinetic Variables

Peak impact force ASI% for the Mobile-10MWT and Immobile-10MWT in the Low-risk and High-risk groups were 6.77 ± 4.78 % vs. 6.55 ± 4.57 %; and 6.45 ± 2.61 % vs. 7.46 ± 7.95 %, respectively. There was no significant test*falls-risk group interaction for peak impact force ($F_{(1, 31)}$ = 0.02, p = 0.893) and no main effects of test ($F_{(1, 31)}$ = 0.22, p = 0.642) or falls-risk group ($F_{(1, 31)}$ = 0.04, p = 0.844). Left and right foot peak impact force ($N \cdot BW^{-1}$) data for each test are shown in Table 8.

Table 8. *Mean ± SD of peak impact force ($N \cdot BW^{-1}$) for the left and right foot during the Mobile-10MWT and Immobile-10MWT*

Test	Low-Risk		High-Risk	
	Left	Right	Left	Right
Mobile-10MWT	1.00 ± 0.14	1.03 ± 0.13	1.01 ± 0.17	1.01 ± 0.16
Immobile-10MWT	1.01 ± 0.14	1.02 ± 0.13	1.02 ± 0.19	1.01 ± 0.18

Average loading rate ASI% for the Mobile-10MWT and Immobile-10MWT in the Low-risk and High-risk groups were 9.71 ± 7.64 % vs. 18.94 ± 11.30 %; and 11.84 ± 6.73 % vs. 20.11 ± 14.72 %, respectively (Figure 14). There was no significant test*falls-risk group interaction ($F_{(1, 31)}$ = 0.05, p = 0.819) and no main effect of test ($F_{(1, 31)}$ = 0.63, p = 0.433). However, there was a significant main effect of falls-risk group ($F_{(1, 31)}$ = 6.06, p = 0.020), but no post-hoc differences between Mobile-10MWT Low-risk vs. High-risk (-9.23 ± 4.12, p = 0.056) or Immobile-10MWT Low-risk vs. High-risk (-8.27 ± 4.12, p = 0.096).

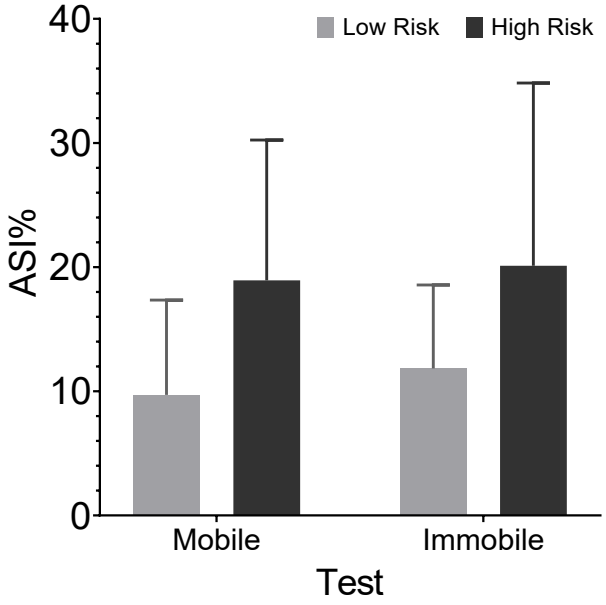


Figure 14*. Mean ± SD of average loading rate ASI% for the left and right foot during the Mobile-10MWT and Immobile-10MWT*

Although average loading rate ASI% differed by falls-risk group overall, this effect was not specific to either test condition. The corresponding left and right foot average loading rate ($N \cdot BW^{-1} \cdot s^{-1}$) values for each test are presented in Table 9.

Table 9. *Mean ± SD of average loading rate ($N \cdot BW^{-1} \cdot s^{-1}$) for the left and right foot during the Mobile-10MWT and Immobile-10MWT*

Test	Low-Risk		High-Risk	
	Left	Right	Left	Right
Mobile-10MWT	8.46 ± 2.04	8.35 ± 2.16	9.07 ± 3.63	8.43 ± 4.07
Immobile-10MWT	8.93 ± 2.10	8.38 ± 1.76	8.57 ± 3.46	8.40 ± 3.63

Midstance force ASI% for the Mobile-10MWT and Immobile-10MWT in the Low-risk and High-risk groups were 5.56 ± 4.93 % vs. 6.67 ± 4.20 %; and 6.44 ± 5.29 % vs. 10.20 ± 10.23 %, respectively. There was no significant test*falls-risk group interaction for midstance force ($F_{(1, 31)}$ = 0.66, p = 0.422) and no main effects of test ($F_{(1, 31)}$ = 1.84, p = 0.185) or falls-risk group ($F_{(1, 31)}$ = 1.60, p = 0.216). Left and right foot midstance force ($N \cdot BW^{-1}$) data for each test are shown in Table 10.

Table 10. *Mean ± SD of midstance force ($N \cdot BW^{-1}$) for the left and right foot during the Mobile-10MWT and Immobile-10MWT*

Test	Low-Risk		High-Risk	
	Left	Right	Left	Right
Mobile	0.81 ± 0.11	0.82 ± 0.11	0.85 ± 0.10	0.83 ± 0.09
Immobile	0.81 ± 0.10	0.82 ± 0.12	0.86 ± 0.10	0.83 ± 0.06

Peak active force ASI% for the Mobile-10MWT and Immobile-10MWT in the Low-risk and High-risk groups were 8.10 ± 4.31 % vs. 9.52 ± 6.05 %; and 7.63 ± 4.52 % vs. 10.26 ± 6.29 %, respectively. There was no significant test*falls-risk group interaction for peak active force ($F_{(1, 31)}$ = 0.87, p = 0.358) and no main effects of test ($F_{(1, 31)}$ = 0.05, p = 0.833) or falls-risk group ($F_{(1, 31)}$ = 1.11, p = 0.301). Left and right foot peak active force ($N \cdot BW^{-1}$) data for each test are shown in Table 11.

Table 11. *Mean ± SD of peak active force ($N \cdot BW^{-1}$) for the left and right foot during the Mobile-10MWT and Immobile-10MWT*

Test	Low-Risk		High-Risk	
	Left	Right	Left	Right
Mobile	1.15 ± 0.15	1.15 ± 0.14	1.16 ± 0.17	1.12 ± 0.12
Immobile	1.12 ± 0.16	1.14 ± 0.16	1.17 ± 0.16	1.12 ± 0.10

Impulse ASI% for the Mobile-10MWT and Immobile-10MWT in the Low-risk and High-risk groups were 6.92 ± 2.55 % vs. 8.26 ± 6.45 %; and 6.45 ± 3.54 % vs. 8.73 ± 5.90 %, respectively. There was no significant test*falls-risk group interaction for impulse ($F_{(1, 31)}$ = 0.69, p = 0.414) and no main effects of test ($F_{(1, 31)}$ = 0.00, p > 0.9999) or falls-risk group ($F_{(1, 31)}$ = 0.98, p = 0.33). Left and right foot impulse ($N \cdot s \cdot BW^{-1}$) data for each test are shown in Table 12.

Table 12. *Mean ± SD of impulse ($N \cdot s \cdot BW^{-1}$) for the left and right foot during the Mobile-10MWT and Immobile-10MWT*

Test	Low-Risk		High-Risk	
	Left	**Right**	**Left**	**Right**
Mobile-10MWT	0.60 ± 0.09	0.62 ± 0.09	0.62 ± 0.09	0.60 ± 0.07
Immobile-10MWT	0.58 ± 0.08	0.59 ± 0.09	0.62 ± 0.09	0.60 ± 0.06

5.4 Predictors of Falls-Risk Classification (Binary Logistic Regression).

A binary logistic regression was performed to examine whether selected dynamic asymmetry variables predicted static balance-derived falls-risk classification. Falls-risk status was coded as Low-risk = 0 and High-risk = 1. The predictor set included mean swing time ASI% and mean average loading rate ASI%. These variables were selected because they differed between falls-risk groups and were considered sufficiently independent for inclusion in the model.

The overall model was significant, indicating that the predictor set improved classification relative to the null model, $\chi^2(2) = 12.393$, $p = 0.002$. Model fit was acceptable, with a Hosmer-Lemeshow test of $\chi^2(8) = 6.305$, $p = 0.613$, indicating good correspondence between predicted and observed classifications. The model accounted for 42.9% of the variance in falls-risk classification according to the Nagelkerke R^2 ($R^2 = 0.429$).

The fitted model was:

$$\text{logit}(p) = -2.454 + 0.216(T_{sw}_ASI_{\%}) + 0.129(AVLR_ASI_{\%})$$

where p represents the probability of being classified as High-risk, T_{sw}_$ASI_{\%}$ represents the mean swing time absolute symmetry index, and AVLR_$ASI_{\%}$ represents mean average loading rate absolute symmetry index.

Overall classification accuracy was 75.8%, with the model correctly identifying 66.7% of Low-risk participants and 81.0% of High-risk participants. These findings show that the predictor set was useful for predicting static balance-derived falls-risk classification, with greater accuracy for High-risk than Low-risk participants.

Chapter Six: Discussion

The aim of this study was to examine the relationships among questionnaire-based falls-risk screening, static balance, and dynamic balance in community-dwelling older adults. Specifically, the study compared STEADI screening classification with a falls-risk classification derived from static balance assessment, examined differences in static centre-of-pressure outcomes and dynamic gait asymmetry between falls-risk groups, compared gait asymmetry following periods of mobility and immobility, and investigated whether selected dynamic gait-asymmetry variables predicted static balance-derived falls-risk classification. It was hypothesised that STEADI would demonstrate limited agreement with the static balance-derived classification, with the objective assessment identifying some participants with elevated postural sway who were classified as Low-risk by STEADI. It was also hypothesised that static and dynamic balance measures would be related but not interchangeable, such that static balance-derived classification would be associated with selected dynamic gait-asymmetry outcomes without accounting for performance across all measured variables. Finally, gait asymmetry was expected to be greater following the period of immobility and among participants classified as High-risk by the static balance assessment.

Several key findings emerged from the present study. First, static balance-derived classification identified substantially more participants as High-risk than STEADI, and agreement between the two methods was low. This demonstrated that questionnaire-based screening and objective static balance assessment did not classify falls-risk consistently within the present cohort. Second, participants classified as High-risk by the static balance assessment demonstrated greater centre-of-pressure movement across multiple outcomes, including path length, velocity, frequency, spatial dispersion, and anterior-posterior sway regularity, indicating clear differences in quiet-standing postural control between groups. Third, selected dynamic asymmetry variables differed between the static balance-derived Low- and High-risk groups. These differences were predominantly observed in spatiotemporal measures, while the average loading rate was the only kinetic variable to demonstrate a significant falls-risk group effect. Fourth, dynamic asymmetry did not differ significantly between the Mobile-10MWT and Immobile-10MWT conditions, indicating that the brief period of immobility did not produce the expected increase in gait asymmetry

under the conditions tested. Finally, mean swing time ASI% and mean average loading rate ASI% jointly contributed to a significant logistic regression model that accounted for 42.9% of the variation in static balance-derived falls-risk classification.

Overall, the findings supported the first and second hypotheses but only partial support for the third. The slight agreement between STEADI and the static balance-derived classification, together with the greater number of participants identified as High-risk by the objective static assessment, supported the hypothesis that some participants with elevated postural sway would not be identified as High-risk by STEADI. The selective differences in dynamic gait asymmetry between static balance-derived groups, collectively with the significant regression model, also supported the hypothesis that static and dynamic balance measures would be related but not interchangeable. However, gait asymmetry did not differ significantly between the Mobile-10MWT and Immobile-10MWT conditions, meaning that the expected effect of immobility was not supported. Although several asymmetry variables were greater in the High-risk group, these differences were not evident across all outcomes. Collectively, the findings indicate that questionnaire screening, static balance, and dynamic gait asymmetry provide related but non-equivalent information about falls-related balance performance in older adults. The following sections discuss these findings in relation to previous literature, beginning with STEADI and static balance-derived falls-risk classification.

6.1 Falls-Risk classification

6.1.1 STEADI

The STEADI Stay Independent questionnaire classified 27 participants (81.8%) as Low-risk and six participants (18.2%) as High-risk. This distribution was similar to that reported by Eckstrom et al. (2017), who implemented the questionnaire among 773 adults aged 65 years and over attending academic primary-care clinics in Oregon, United States. Using the same threshold of four or more points, 78% of participants were classified as Low-risk and 22% as High-risk. Although the populations were recruited through different settings, the similar proportions indicate that both samples were predominantly classified as Low-risk using the Stay Independent questionnaire. In contrast, Nithman and Vincenzo (2019) reported that 46.8% of their sample, drawn from community and retirement-facility settings in two regions of the United States, met the at-risk threshold. The higher proportion identified in that study

may partly reflect differences in participant setting and functional characteristics, including the older age of the retirement-facility group and the high prevalence of previous falls reported within the community sample. The findings should therefore be interpreted in relation to the characteristics and recruitment context of each population.

A limitation of STEADI is its reliance on the accuracy of recollection. Cummings et al. (1988) found that 13-32% of older adults with prospectively recorded falls did not recall those events when questioned retrospectively over periods ranging from three to 12 months. Ganz et al. (2005) similarly concluded that retrospective questioning is less sensitive than prospective monitoring using regular falls calendars or interviews. Older adults may also conceptualise falls differently, describing events as slips, trips, stumbles, or losses of balance rather than identifying them as falls (Zecevic et al., 2006). Consequently, the absence of a reported fall does not necessarily establish that no fall occurred. Furthermore, self-perception and the social meaning attached to falls may influence responses to the questionnaire (Dollard et al., 2012). Some older adults may not want others to perceive them as being at risk because falls-related labels may be associated with frailty, dependence, or a threat to independence (McInnes & Askie, 2004; Yardley et al., 2006). This may be particularly relevant in independently mobile older adults who do not perceive themselves as vulnerable despite emerging changes in balance performance.

The predominance of Low-risk classifications in the present study should be considered in relation to STEADI's intended role. STEADI was developed as a structured clinical pathway (Phelan et al., 2015; Stevens & Phelan, 2013), whereby individuals identified as being at risk progress to more detailed assessment of modifiable contributors to falls (Montero-Odasso et al., 2022; Phelan et al., 2015). In contrast, individuals classified as Low-risk generally receive primary-prevention guidance, including education and recommendations for strength and balance activities, rather than progressing to the same detailed assessment (Phelan et al., 2015; Stevens & Phelan, 2013). Consequently, the possibility that an individual with underlying falls-related vulnerability may be classified as Low-risk is important within this staged pathway. Nithman and Vincenzo (2019) demonstrated this concern in community-dwelling older adults, reporting that 57% of those classified as Low-risk by the STEADI algorithm had experienced a fall during the previous 12 months. These findings indicate that a Low-risk STEADI classification does not necessarily reflect the absence of falls-related

vulnerability, and that some individuals who remain vulnerable to falls may therefore not progress to more detailed assessment or targeted risk-management strategies.

The large proportion classified as Low-risk should not be interpreted as confirmation that these participants had no balance impairment. It indicates only that they did not meet the questionnaire threshold for an at-risk classification based on the information reported. The limitations of recall, self-perception, disclosure, and screening sensitivity provide a rationale for considering objective physical assessment alongside questionnaire-based screening, particularly in independently mobile older adults whose balance deficits may not yet have resulted in recognised or reported falls. The static balance assessment, therefore, provided an objective means of determining whether the predominantly Low-risk STEADI classifications were reflected in participants' quiet-standing balance performance.

6.1.2 Static Balance

The static balance assessment classified 12 participants as Low-risk, three as Moderate-risk, and 18 as High-risk. For subsequent binary analyses, the Moderate- and High-risk categories were combined, resulting in 21 participants in the High-risk group and 12 in the Low-risk group. As only three participants were classified as Moderate-risk, the combined group predominantly reflected the original High-risk classifications, which accounted for 85.7% of the group, compared with 14.3% from the Moderate-risk category.

The assessment used the centre-of-pressure trajectory recorded in the anterior-posterior and mediolateral directions during quiet standing to derive measures, including path length, velocity, mean frequency, and the area of the 95% ellipse. These measures describe the amount, rate, and spatial extent of centre-of-pressure movement and have been associated with postural instability and falls-related outcomes in older adults (Howcroft et al., 2017; Maki et al., 1994; Pajala et al., 2008; Prieto et al., 1996). The predominance of High-risk classification, therefore, indicates that many participants demonstrated elevated postural sway during the eyes-closed standing assessment.

A limitation of the static classification is that it reflects balance performance during one testing session and under one standardised condition. Although force-platform centre-of-pressure measures generally demonstrate acceptable test-retest reliability, some within-person variability and familiarisation effects can occur between sessions (Goble et al., 2021;

Harro & Garascia, 2019). Performance may also be influenced temporarily by factors such as fatigue (Inoue et al., 2013), attentional demands (Manor et al., 2010), sleep loss and associated lapses in attention (Patel et al., 2008), and medication use (Robin et al., 1996; Tsunoda et al., 2010), all of which may alter postural-control performance at the time of testing. The classification should therefore be interpreted as an assessment of balance performance at that time rather than as a fixed characteristic across all occasions. However, the result was averaged across three recorded trials, reducing the influence of a single atypical performance and providing a more representative estimate of postural sway than a single trial alone (Golriz et al., 2012). Nevertheless, under the standardised conditions assessed, those classified as High-risk demonstrated greater postural sway and poorer quiet-standing steadiness. The contrasting outcomes of the two classification approaches, therefore, highlighted the need to examine their level of agreement in the present cohort.

6.2 Comparing STEADI vs. Static Balance Risk Classification

The STEADI and static balance-derived classifications showed substantial directional disagreement within the present cohort. McNemar's test indicated that the classifications differed significantly in direction, while Cohen's kappa showed only slight agreement in how the two methods classified individual participants. The direction of disagreement was particularly notable. Sixteen participants, representing 48.5% of the cohort, were classified as Low-risk by STEADI but High-risk by the binary static balance classification. In contrast, only one participant, representing 3.0% of the cohort, was classified as High-risk by STEADI but Low-risk by the static assessment. Thus, disagreement occurred predominantly because participants who did not reach the STEADI at-risk threshold demonstrated elevated postural sway during objective static balance testing. This pattern is consistent with previous research showing that questionnaire-based falls-risk tools do not identify all older adults with falls-related vulnerability, particularly when classification depends on recalled or recognised risk indicators (Burns et al., 2022; Nithman & Vincenzo, 2019).

The single participant classified as High-risk by STEADI, but Low-risk by the static assessment, may have had a falls-related vulnerability that was not reflected in quiet-standing performance under the standardised conditions assessed. This isolated pattern contrasts with the much larger group showing the opposite result, indicating that reliance on

STEADI alone would not have identified nearly half of the cohort as demonstrating elevated static postural sway. Although STEADI may have accurately reflected the absence of sufficient reported indicators, those participants nevertheless showed poorer quiet-standing steadiness under the standardised conditions assessed. This finding is important because greater force-plate-measured postural sway has been associated with falls history and prospective falls in older adults, while normative centre-of-pressure data can be used to identify sway performance that is atypical relative to that of comparable adults (Goble & Baweja, 2018; Howcroft et al., 2017; Maki et al., 1994; Pajala et al., 2008).

These findings support the first hypothesis that STEADI would demonstrate limited agreement with the static balance-derived classification and that objective assessment would identify elevated postural sway among participants classified as Low-risk by STEADI. They also have implications for staged falls-risk pathways in which further physical assessment depends on the initial questionnaire result. Within the present cohort, using STEADI as the sole gateway to objective assessment would have left a substantial proportion of participants with elevated static postural sway unidentified. Incorporating an objective static balance assessment alongside questionnaire screening may therefore strengthen early identification by detecting postural-control deficits that are not recognised, recalled, or reported through STEADI. The marked disagreement between classifications, therefore, required closer examination of the static balance characteristics underlying the Low- and High-risk groups. The following section considers whether the High-risk classification was supported by a consistent pattern across the centre-of-pressure measures

6.3 Static Balance Characteristics of the Low- and High-Risk Groups

When participants were separated according to the static balance-derived classification, the High-risk group demonstrated a consistent pattern of greater centre-of-pressure movement across several related outcomes. Mean path length was almost twice as large in the High-risk group as in the Low-risk group (Figure 7A), while mean velocity was also substantially higher (Figure 7B). The High-risk group additionally demonstrated a larger 95% ellipse area (Figure 8D). Collectively, these findings indicate that the centre-of-pressure travelled farther, moved more rapidly, and occupied a broader area while participants maintained quiet standing without visual input. Furthermore, because all trials were of equal duration, the longer path

length and higher velocity show that a greater amount of postural movement occurred at a faster average rate, while the larger ellipse area demonstrates that this movement was distributed across a wider region beneath the feet (Prieto et al., 1996; Quijoux et al., 2021). Greater and faster centre-of-pressure movement may indicate that more continuous postural regulation was required to maintain the centre-of-mass relative to the base of support (Quijoux et al., 2021; Winter, 1995). Although these adjustments were sufficient to maintain standing during the assessment, the greater requirement for postural regulation may become more consequential when attentional demands increase, or sensory information is reduced or altered, as these conditions can further challenge postural control in older adults (Manchester et al., 1989; Manor et al., 2010; Teasdale et al., 1991).

The present findings are consistent with previous research linking greater centre-of-pressure movement with falls-related outcomes in older adults. Wiśniowska-Szurlej et al. (2022) reported greater eyes-closed centre-of-pressure path length among older adults identified as having elevated falls-risk than among those without elevated risk. Pajala et al. (2008) similarly found that community-dwelling older women with greater centre-of-pressure movement had a higher prospective risk of indoor falls. Howcroft et al. (2017) also reported higher centre-of-pressure velocity and other sway measures among prospective multiple fallers than non-fallers during eyes-closed standing. These studies support the interpretation that greater centre-of-pressure movement is relevant to falls-related postural instability.

In addition to the longer path length and higher velocity, the High-risk group also demonstrated a higher mean frequency than the Low-risk group (Figure 7C). Mean frequency estimates the theoretical rotational frequency of the centre-of-pressure trajectory by relating mean velocity to the average distance of the trajectory from its mean position (Prieto et al., 1996; Quijoux et al., 2021). The higher value, therefore, indicates that centre-of-pressure movement occurred at a greater rate relative to the trajectory's spatial scale. When considered alongside the longer path length and higher velocity, this suggests that the High-risk group required faster and more continuous regulation of centre-of-pressure movement to maintain an upright stance. Prieto et al. (1996) identified mean frequency as one of several hybrid centre-of-pressure measures that showed stronger age-related differences during eyes-closed than eyes-open standing. However, because evidence directly linking mean frequency with prospective falls remains limited, the present result is best

interpreted as part of the broader centre-of-pressure profile rather than as an independent indicator of future falls.

The spatial outcomes further showed that the High-risk group's centre-of-pressure movement extended more widely in both the anterior-posterior and mediolateral directions. The High-risk group demonstrated a larger 95% ellipse area than the Low-risk group (Figure 8D), together with greater anterior-posterior and mediolateral root-mean-square displacement (Figures 8B and 8C). Anterior-posterior range, mediolateral range, and mean distance from the average centre-of-pressure position were also greater in the High-risk group (Figures 9A-C). Collectively, these findings indicate that the High-risk group's centre-of-pressure trajectory occupied a broader overall area, showed greater typical displacement from the mean position, and reached larger outer excursions in both directions (Prieto et al., 1996; Quijoux et al., 2021). This interpretation is consistent with previous evidence linking directional sway with falls-related outcomes. Maki et al. (1994), for example, found that a twofold increase in mediolateral root-mean-square displacement during blindfolded standing was associated with greater odds of future falls among independently living older adults, while Howcroft et al. (2017) identified anterior-posterior range and root-mean-square measures among variables contributing to prospective faller-classification models. The directional differences may also provide insight into the mechanisms commonly used to regulate quiet standing. Anterior-posterior centre-of-pressure movement is controlled predominantly through ankle plantar-flexor and dorsiflexor activity, whereas mediolateral control depends more strongly on coordinated loading and unloading between the limbs through the hip abductors and adductors (Winter, 1995). Greater root-mean-square displacement and range in both directions, therefore, indicate that the High-risk group demonstrated larger excursions within directions commonly regulated through both ankle-mediated and hip-mediated mechanisms. However, because joint motion, muscle activity, and the individual contribution of each foot were not measured, the findings cannot establish that either strategy was specifically impaired.

The temporal organisation of centre-of-pressure movement also differed between groups. Anterior-posterior approximate entropy was higher in the High-risk group than in the Low-risk group (Figure 8A), however mediolateral approximate entropy did not differ significantly. Approximate entropy reflects the extent to which short sequences within a time series

repeat as the signal progresses, with higher values indicating fewer repeated patterns and a more irregular or less predictable temporal structure (Fino et al., 2016). Accordingly, the High-risk group demonstrated both greater anterior-posterior sway magnitude, reflected in the higher root-mean-square displacement and range, and a less repetitive anterior-posterior sway pattern. In contrast, mediolateral sway magnitude was greater in the High-risk group, while its temporal regularity was comparable between groups. This distinction indicates that sway magnitude and temporal structure captured different features of centre-of-pressure behaviour. Given that entropy is sensitive to calculation method, signal direction, and testing condition, the higher anterior-posterior approximate entropy should not be interpreted independently as evidence of poorer balance (Fino et al., 2016). Rather, when considered alongside the spatial findings, it indicates that the High-risk group differed in both the magnitude and temporal patterning of anterior-posterior postural sway.

The eyes-closed condition provides important context for the overall pattern because removing visual input increases reliance on somatosensory and vestibular information for maintaining upright stance (Manchester et al., 1989; Peterka, 2018; Teasdale et al., 1991). Under this standardised condition, the High-risk group demonstrated greater accumulated movement, faster movement, broader spatial excursions, and altered anterior-posterior regularity. The assessment was not designed to identify which sensory or motor system contributed to these differences, but it demonstrated a coherent difference in integrated postural performance between groups. Although path length was central to the static classification, the accompanying differences in velocity, frequency, ellipse area, directional displacement, and approximate entropy show that the distinction between groups was not limited to a single centre-of-pressure outcome. Collectively, these findings support the interpretation that participants classified as High-risk demonstrated poorer quiet-standing steadiness under the standardised eyes-closed condition.

Static standing nevertheless represents only one context in which balance must be controlled. Falls frequently occur during walking, turning, transfers, and other activities in which the centre-of-mass and base of support are continuously changing. Recent research has therefore examined between-limb asymmetry during gait as an additional indicator of dynamic balance and falls-related vulnerability, including asymmetry in loading and temporal variables measured using in-shoe sensors (Nakanowatari, Hoshi, Asao, et al., 2024;

Nakanowatari, Hoshi, Sone, et al., 2024). The following section examines whether the static balance-derived Low- and High-risk groups also differed in absolute symmetry index outcomes during the Mobile-10MWT and Immobile-10MWT.

6.1 Dynamic Balance

6.1.1 Effect of Mobility and Immobility on Gait Asymmetry

Gait asymmetry did not differ between the Mobile-10MWT and Immobile-10MWT for any spatiotemporal or kinetic variable. No test-condition-by-falls-risk-group interactions were identified, indicating that the static balance-derived Low- and High-risk groups responded similarly to the two walking conditions. Therefore, the one-minute seated period did not produce the expected increase in between-limb asymmetry during the subsequent 10-metre walk, and the condition-related component of the third hypothesis was not supported.

One explanation is that one minute of sitting provided an insufficient period of immobility to produce a measurable change in subsequent gait asymmetry. The one-minute duration was selected to provide a standardised comparison with the one-minute walking period preceding the Mobile-10MWT. Matching the duration of the preparatory conditions allowed the study to examine whether gait asymmetry differed depending on whether participants had been mobile or immobile immediately before the 10-metre walk, while minimising differences in the time preceding each assessment. One explanation is that one minute of sitting was insufficient to produce a measurable change in subsequent gait asymmetry. This comparison was physiologically plausible because transitioning to standing requires rapid cardiovascular adjustment to limit the fall in blood pressure associated with postural change, and initial orthostatic hypotension is common among older adults (Tran et al., 2021). Slower blood-pressure recovery after standing has also been independently associated with poorer gait performance in community-dwelling older adults (Briggs et al., 2020). In addition, prolonged uninterrupted sitting can reduce lower-limb vascular shear rate and impair endothelial function, although the relevant experimental studies used seated periods of three hours and examined vascular outcomes rather than subsequent gait asymmetry (Morishima et al., 2016; Thosar et al., 2015; Vranish et al., 2017). The available evidence therefore supports examining walking after seated inactivity, but does not establish that one minute of sitting is sufficient to alter gait asymmetry. The absence of a condition effect in the

present study suggests that any cardiovascular, circulatory, postural, or neuromotor adjustments associated with the seated condition were either too small, too brief, or not reflected in the asymmetry measures obtained during the subsequent 10-metre walk.

Another consideration is the direct transition from sitting to walking required during the Immobile-10MWT. Sit-to-walk requires the centre-of-mass to be transferred forwards and upwards while the base of support changes from the chair and both feet to alternating single-limb support, placing substantial demands on postural control during seat-off and gait initiation (Åberg et al., 2010; Buckley et al., 2009). Older adults may manage these demands by adopting more cautious movement strategies. Buckley et al. (2009) found that healthy older adults generated less forward momentum before rising, delayed gait initiation until they had reached a more upright posture and took shorter and slower initial steps than younger adults. Chen and Chou (2013) similarly found that older adults, particularly those with a history of falling, favoured stability over forward progression during sit-to-walk by generating less forward centre-of-mass velocity and directing a greater proportion of momentum upwards. In the present study, data collection commenced from the first walking step following the seated period, and no increase in spatiotemporal or kinetic asymmetry was observed between the Mobile-10MWT and Immobile-10MWT conditions.

The relatively simple nature of the walking task may also have contributed to this finding. Participants walked at a self-selected pace along a straight, level pathway without additional cognitive or environmental demands. Although gait requires attentional control, these demands increase when walking must be adapted or performed concurrently with another task (Woollacott & Shumway-Cook, 2002; Yogev-Seligmann et al., 2008). The controlled conditions of the present assessment may have allowed participants sufficient opportunity to stabilise their gait following the seated transition. Any physiological, postural, or neuromotor adjustments associated with the seated period and subsequent transition to walking may consequently have resolved before producing a detectable difference in asymmetry across the 10-metre walk. Future research could examine whether placing greater demands on gait initiation and adaptation produces a more pronounced effect.

Importantly, the absence of a condition effect does not indicate that gait was symmetrical. It shows only that the magnitude of asymmetry was not increased by the one-minute seated period. Differences between the static balance-derived Low- and High-risk groups were

nevertheless evident when the two walking conditions were considered together. The absence of an interaction indicates that there was no evidence that the magnitude of these group differences depended on whether walking followed mobility or immobility. The observed asymmetry therefore appeared to reflect more persistent differences in gait control between the groups rather than a temporary response to sitting. The implications of these group differences for dynamic balance and falls-related vulnerability are examined in the following section.

6.1.2 Gait Asymmetry Between Static Balance-Derived Falls-Risk Groups

Although gait asymmetry did not differ between the Mobile-10MWT and Immobile-10MWT conditions, significant main effects of static balance-derived falls-risk group were identified for selected variables when the two walking conditions were considered together. When the two walking conditions were considered together, the High-risk group demonstrated greater ASI% for ground contact time, swing time, peak impact time, midstance time, and average loading rate. In contrast, stride duration, stride frequency, time to active peak force, peak impact force, midstance force, peak active force, and impulse did not differ between groups. The group differences were therefore concentrated mainly in gait timing, with average loading rate representing the only kinetic measure to differ between groups. This pattern indicates that poorer static postural control was associated with selected differences in the between-limb timing of support and limb advancement, as well as the timing and rate of weight acceptance, rather than a generalised increase in asymmetry across the entire gait cycle. The following sections examine the spatiotemporal and kinetic findings separately.

6.1.2.1 Spatiotemporal Asymmetry

The High-risk group demonstrated greater ASI% for ground contact time, swing time, peak impact time, and midstance time when the two walking conditions were considered together (Figures 10-13). The largest between-group differences were observed for peak impact time, with mean ASI% values of 12.24% and 13.04% in the High-risk group compared with 4.68% and 5.82% in the Low-risk group during the Mobile-10MWT and Immobile-10MWT, respectively. Swing time showed a similar pattern, with corresponding values of 10.43% and 8.35% in the High-risk group and 3.87% and 3.24% in the Low-risk group. Ground contact time and midstance time ASI% were also numerically higher in the High-risk group under both conditions.

Ground contact and swing time reflect the stance and limb-advancement phases, while peak impact time and midstance time describe the timing of early- and mid-stance loading events (Perry & Burnfield, 2010). The greater ASI% values across these measures indicate that the High-risk group demonstrated less symmetrical timing of support, limb advancement, and loading progression through stance. During symmetrical gait, the limbs alternate through broadly comparable periods of support and advancement, allowing the centre-of-mass to be transferred rhythmically from one supporting limb to the other as the base of support changes between successive steps (Perry & Burnfield, 2010; Winter, 1995). Greater asymmetry means that one limb remains in support, advances through swing, or progresses through early stance at a different rate from the contralateral limb. Consequently, each side may provide a different temporal contribution to forward progression and body support (Perry & Burnfield, 2010).

Several interacting physical and cognitive factors may contribute to this temporal asymmetry. Age-related reductions in lower-limb strength and muscle function may limit the capacity to support and control the body during walking, while declines in proprioceptive and broader somatosensory function may reduce the accuracy of information concerning limb position and movement (Shaffer & Harrison, 2007; Sturnieks et al., 2008). Jia et al. (2024) similarly proposed lower-limb strength, proprioceptive function, motor planning, and cognitive processing as possible contributors to altered stance timing in older adults. Collectively, these factors could influence how consistently support and limb advancement are organised between sides. Greater between-limb asymmetry may therefore reflect underlying differences in physical or sensorimotor capacity, or adaptations used to maintain forward progression despite those differences. Alongside these physical and sensorimotor influences, the cognitive demands of gait may also affect the consistency of temporal control. Although uncomplicated walking is relatively rhythmical and automatic, attention and executive function contribute increasingly when gait must be monitored, adapted, or performed concurrently with another task (Yogev-Seligmann et al., 2008). Springer et al. (2006) found that dual-tasking increased gait variability in older adults with a history of falling, but not in older non-fallers, and that greater dual-task-related gait variability was associated with poorer executive function. This suggests that some older adults with falls-

related vulnerability may have fewer cognitive resources available for maintaining gait consistency when attentional demands increase.

Physical and cognitive limitations may therefore contribute to unequal timing between limbs, while the functional importance of this pattern may become more apparent when walking demands increase. If one limb remains in support for longer or requires more time to advance, weight transfer and preparation for the subsequent foot contact occur over different time periods on each side (Perry & Burnfield, 2010). Rapid responses to an unexpected balance disturbance also require the timely selection and execution of an appropriate recovery step, and these reactions can be impaired with ageing (Maki & McIlroy, 2006). Accordingly, the greater temporal asymmetry observed in the High-risk group may indicate less consistent organisation of support and advancement that could become more consequential when rapid gait adaptation or balance recovery is required.

Previous research supports the relevance of these temporal characteristics to falls-related assessment. Jia et al. (2024) found that a signed left-right difference in standing time contributed to a fall-history prediction model in nursing-home residents aged over 80 years, indicating that between-limb differences in support timing contained information relevant to falls-related status. Their reported calculation retained the direction of the difference between limbs, whereas the present study used ASI% to quantify asymmetry magnitude irrespective of whether the longer duration occurred in the left or right limb. This approach was appropriate because the limb producing the longer time differed among participants. Averaging signed values would have allowed positive and negative differences to cancel at the group level, potentially indicating little overall asymmetry despite substantial between-limb differences within individual participants. ASI% therefore enabled comparison of the magnitude of asymmetry independently of whether the left or right limb produced the larger value.

The importance of temporal regulation is also supported by prospective evidence. Among 102 community-dwelling older adults, Marques et al. (2018) reported that stance-time variability, swing time, and stride length each demonstrated greater than 70% sensitivity for identifying participants who subsequently fell. Fallers also demonstrated longer stance time, shorter swing time, and greater stance-time variability. Although variability describes fluctuations across successive strides and ASI% describes the difference between limbs, both

indicate reduced consistency in the temporal organisation of gait. Marques et al. (2018) attributed these alterations partly to age-related neuromuscular impairments that may reduce the regulation of postural control during walking. When considered alongside the present findings, this suggests that impaired temporal control may be expressed either as inconsistent timing across strides or as unequal timing between limbs, both of which may complicate the continuous regulation of balance during forward progression.

The present findings add to this evidence by showing that older adults with poorer quiet-standing balance also demonstrated greater between-limb asymmetry in selected temporal gait events. This association suggests that the postural-control differences evident during static standing were reflected partly in how the limbs alternated support, advancement, and loading during walking. Importantly, this relationship was selective rather than evident across every spatiotemporal outcome. The accompanying difference in average loading rate provides further insight into this process by showing that the High-risk group also differed in how rapidly the two limbs accepted force during early stance. This kinetic finding is considered in the following section.

6.1.2.2 Kinetic Asymmetry

Average loading rate was the only kinetic variable to differ between the static balance-derived falls-risk groups. Mean ASI% values were 18.94% and 20.11% in the High-risk group compared with 9.71% and 11.84% in the Low-risk group during the Mobile-10MWT and Immobile-10MWT, respectively (Figure 14). Average loading rate reflects how rapidly vertical force increases following initial contact as body weight is accepted onto the supporting limb (Perry & Burnfield, 2010; Renner et al., 2019). In the present study, it was calculated as the slope of the force-time curve between 20% and 80% of the first impact peak. The greater ASI% therefore indicates that the rate of force increase during early weight acceptance differed more between limbs in the High-risk group.

This difference was specific to average loading rate, as peak impact force, midstance force, peak active force, and impulse ASI% did not differ between groups. The kinetic distinction therefore appeared to relate more to how rapidly force developed during early weight acceptance than to the magnitude or cumulative amount of vertical force. When considered alongside the greater peak impact time and midstance time asymmetry described in the

preceding section, the findings indicate that between-limb differences were present in both the timing of selected stance events and the rate of force development during early stance. These variables describe related features of the force-time profile because average loading rate is influenced partly by the time over which force increases. However, peak impact time and midstance time represent distinct gait events and were not the specific interval used to calculate average loading rate. Their concurrent group differences therefore suggest less symmetrical temporal and kinetic organisation of early stance, rather than demonstrating that asymmetry in one measure produced asymmetry in another (Perry & Burnfield, 2010; Renner et al., 2019). This pattern can be considered alongside previous research examining whether between-limb loading asymmetry is associated with falls-related outcomes in older adults.

Nakanowatari, Hoshi, Sone, et al. (2024) found that peak-force asymmetry measured using in-shoe sensors was greater in older Japanese adult fallers than non-fallers, whereas contact-time, loading-rate, and impulse asymmetry did not differ between the groups. The authors suggested that loading asymmetry may reflect differences between the limbs in motor function and plantar load-receptor input. They also noted that local asymmetry in knee or hip function could contribute to asymmetrical limb motion and recommended further investigation of joint function and pain as potential influences. They further proposed that some peak-force asymmetry may represent normal adaptation, but that excessive asymmetry during a simple walking task could indicate abnormal loading control and provide an early marker of falls-related vulnerability in otherwise mobile older adults. This interpretation is relevant to the present findings because the static balance-derived High-risk group also demonstrated greater between-limb asymmetry in a loading-related variable during simple walking. However, the specific variable differed: average loading-rate asymmetry distinguished the groups in the present study, whereas peak-force asymmetry did not. Thus, both studies suggest that loading asymmetry may contain information relevant to falls-related vulnerability, although the feature of the loading response that differentiated groups was not consistent across studies.

Peak-force asymmetry also demonstrated high test-retest reliability (ICC = 0.942), while the reliability of the remaining asymmetry measures was below 0.75 (Nakanowatari, Hoshi, Sone, et al., 2024). This indicates that peak-force asymmetry was reproduced consistently

across repeated trials, increasing confidence that the measure was less affected by trial-to-trial variation than the other asymmetry outcomes. Conversely, contact-time, loading-rate, and impulse asymmetry were less repeatable across trials, which may have made stable between-group differences in these variables more difficult to detect. This lower repeatability is particularly relevant to the present loading-rate finding because each dynamic condition was represented by a single walking trial. The observed group difference should therefore be interpreted cautiously until the between-trial and between-day reproducibility of average loading-rate ASI% is established in a comparable older population.

In a separate prospective study, Nakanowatari, Hoshi, Asao, et al. (2024) found that both net loading-rate and net impulse asymmetry were greater among Japanese participants who developed frailty during the 10-month follow-up. However, only net impulse asymmetry remained independently associated with frailty development in the multivariable analysis, with each one-percentage-point increase corresponding to 26.9% greater odds of frailty development (adjusted odds ratio = 1.269). Net impulse asymmetry was also associated with peak-force asymmetry in the midfoot region. Nakanowatari, Hoshi, Asao, et al. (2024) proposed that reduced midfoot function, potentially associated with tibialis posterior dysfunction, decreased foot strength, sensitivity or mobility, altered somatosensory function, and age-related plantar deformation could impair the foot's contributions to shock attenuation and propulsion. They suggested that older adults at risk of frailty may rely more on a pull-off than a push-off strategy to generate forward momentum, providing a possible biomechanical explanation for the relationship between loading asymmetry and frailty development.

Although these studies all examined between-limb loading, their findings should be interpreted in relation to the different outcomes used to characterise vulnerability. Peak-force asymmetry distinguished older adults with and without a retrospective fall history in Nakanowatari, Hoshi, Sone, et al. (2024), while net impulse asymmetry prospectively predicted frailty development in Nakanowatari, Hoshi, Asao, et al. (2024). In the present study, average loading-rate asymmetry distinguished groups classified according to quiet-standing postural sway. The contrasting findings may therefore reflect both the different loading characteristics measured and the different outcomes used to define vulnerability. Nakanowatari, Hoshi, Sone, et al. (2024) identified fallers retrospectively from falls reported

during the preceding year, whereas Nakanowatari, Hoshi, Asao, et al. (2024) examined prospective frailty progression over 10 months. The present groups represented differences in an objective sway-based falls-risk classification rather than confirmed past falling or subsequent frailty development. Accordingly, the present findings specifically associate poorer quiet-standing postural control with greater asymmetry in average loading rate. Rather than identifying one kinetic asymmetry measure that consistently characterises vulnerability, the studies suggest that different aspects of between-limb loading may relate to fall history, frailty progression, and sway-based postural vulnerability.

Overall, the dynamic findings support Hypothesis 2 by showing that static balance-derived falls-risk classification was associated with selected gait-asymmetry measures, but not across all gait-asymmetry outcomes. Hypothesis 3 was partially supported because group differences were evident for selected dynamic gait-asymmetry variables, whereas no condition effect was observed. The combined contribution of swing-time and average loading-rate ASI% was therefore examined using logistic regression to determine whether these temporal and kinetic asymmetry measures jointly predicted static balance-derived Low- or High-risk classification.

6.2 Predictors of Falls-Risk Classification

The binary logistic regression showed that mean swing time ASI% and mean average loading rate ASI% jointly differentiated participants classified as Low- or High-risk by the static balance assessment. This suggests that temporal and loading-rate asymmetry captured different features of gait associated with quiet-standing postural control. The weak and non-significant correlation between the predictors ($r = 0.166$, $p = 0.357$) further indicates that there was limited overlap between the two gait-asymmetry measures, supporting their inclusion within the same model.

The two-predictor model significantly improved differentiation between the static balance-derived groups compared with the intercept-only model, $\chi^2(2) = 12.393$, $p = 0.002$. The non-significant Hosmer-Lemeshow test, $\chi^2(8) = 6.305$, $p = 0.613$, provided no evidence of poor model fit, while the Nagelkerke R^2 of 0.429 indicated that the model had a meaningful, although incomplete, explanatory contribution. Overall classification accuracy increased from 63.6% to 75.8% following the inclusion of swing time and average loading-rate

asymmetry. The model correctly classified 81.0% of High-risk participants and 66.7% of Low-risk participants. The higher accuracy for the High-risk group may partly reflect the unequal group sizes, with 21 participants classified as High-risk and 12 as Low-risk. Nevertheless, eight participants overall remained incorrectly classified, showing that the two dynamic measures improved differentiation but did not fully reproduce the static balance-derived classification.

Although the overall model was significant, neither predictor reached statistical significance independently at the conventional threshold. Mean swing time ASI% had a positive coefficient (B = 0.216, p = 0.102, odds ratio = 1.242), as did mean average loading rate ASI% (B = 0.129, p = 0.058, odds ratio = 1.138). After accounting for the other predictor, each one-percentage-point increase in mean swing time ASI% corresponded to an estimated 24.2% increase in the odds of High-risk classification, while each one-percentage-point increase in mean average loading rate ASI% corresponded to an estimated 13.8% increase. However, because the individual coefficients were not statistically significant, these estimates should be interpreted cautiously and do not demonstrate independent effects. Instead, the significant overall model indicates that the two asymmetry measures contributed jointly to differentiation between the static balance-derived groups. This differs from Nakanowatari, Hoshi, Sone, et al. (2024), where peak-force asymmetry distinguished older adults with and without a retrospective fall history, and from Nakanowatari, Hoshi, Asao, et al. (2024), where net impulse asymmetry remained independently associated with prospective frailty development. Therefore, these contrasting findings suggest that the relevance of individual gait-asymmetry measures may depend partly on whether vulnerability is characterised using fall history, subsequent frailty development, or quiet-standing postural sway.

Furthermore, the combined contribution of swing time and average loading rate is consistent with evidence that static and dynamic balance are related only modestly. Kiss et al. (2018) reported a small pooled association between static and dynamic steady-state balance in older adults (r = 0.31), while Rizzato et al. (2021) found no significant relationships between quiet-standing centre-of-pressure measures and dynamic balance performance (r = 0.135-0.202). These findings suggest that performance during quiet standing may overlap with selected aspects of gait control without accounting for dynamic balance more broadly. In the present study, swing time and average loading rate jointly

differentiated the static balance-derived groups, but the remaining misclassification and non-significant individual predictors indicate that this overlap was incomplete.

The regression analysis addressed Aim 4 by examining whether selected dynamic gait-asymmetry variables contributed to static balance-derived falls-risk classification. It also supported Hypothesis 2, which proposed that static and dynamic balance would be related but not interchangeable. The significant model and improved classification accuracy indicate measurable overlap between quiet-standing performance and gait asymmetry, while the remaining misclassification and non-significant individual predictors show that the relationship was incomplete. These findings do not indicate that one assessment approach was more informative than the other, as the static balance classification formed the outcome of the regression and subsequent falls were not monitored. Instead, the results suggest that quiet-standing sway and gait asymmetry provided overlapping but non-equivalent information. Considering both assessment contexts may therefore provide a broader description of falls-related vulnerability than either alone. Their combined use may contribute to a more holistic assessment of falls-risk by accounting for postural control during both standing and walking. The model should therefore be interpreted as exploratory evidence of an association within the present cohort. The limitations relevant to its interpretation are considered in the following section.

6.3 Limitations

The present study was designed to compare static and dynamic balance at a single point in time, rather than prospectively investigate the occurrence of falls. Accordingly, while relationships between assessment-derived measures could be examined, the findings cannot establish whether the observed differences in postural sway and gait asymmetry are associated with subsequent falls. Although this design allowed the assessments to be completed under the same testing conditions, it also meant that the findings reflected each participant's physical and cognitive state on the day of testing rather than their performance across time. This is important because balance and gait are not entirely fixed and may vary in response to temporary changes in sleep, attention, pain, medication effects, fatigue, and recovery status. Experimental sleep deprivation has been shown to impair postural control and adaptation, potentially through temporary lapses in attention (Patel et al., 2008), while

greater attentional demands can also alter postural-control performance in older adults (Shumway-Cook et al., 2000).

Temporary differences in physical condition may similarly affect measured performance. Experimental evidence indicates that lower-limb fatigue can alter gait and postural-control outcomes comparable with those examined in the present study. In healthy older adults, unilateral hip-abductor fatigue increased stride-time variability and reduced mediolateral step-to-step symmetry during walking (Arvin et al., 2015), while experimentally induced lower-limb fatigue altered force-platform centre-of-pressure measures in healthy younger adults (Inoue et al., 2013). However, these protocols were substantially more demanding than the brief assessments used in the present study and do not suggest that the testing protocol itself caused fatigue. Instead, they demonstrate that an individual's level of fatigue or recovery at the time of testing has the potential to influence postural and gait performance. Habitual and recent physical activity may also have contributed further to variation in participants' physical condition at the time of testing. Exercise interventions have been shown to improve static and dynamic balance in older adults, while physical activity level has been associated with maximal gait speed (Aranda-García et al., 2015; Papalia et al., 2020). Differences in participants' usual weekly activity may therefore have contributed to variation in balance and walking performance. In addition, recent fatiguing activity has been shown to alter postural sway, lower-limb strength, reaction time, and falls-risk measures in older adults, indicating that the amount or intensity of activity completed before testing could have influenced fatigue or recovery status (Morrison et al., 2016). Neither habitual physical activity nor recent activity exposure was formally measured in the present study.

Pain and medication use may represent further sources of participant-level variation. Pain has been associated with recurrent falls among community-dwelling older adults (Stubbs et al., 2014), while several medication classes have been associated with increased falls-risk in older adults (Blain et al., 2018). These studies do not establish that pain or medication use altered performance during the present assessments, or that their effects varied between testing days. However, they identify relevant participant-level factors that were not formally assessed and could have differed according to each participant's condition on the day of testing.

Although these factors were not shown to have influenced the present results, they may have contributed to variation in performance. The static balance classifications and gait-asymmetry values should therefore be interpreted as single-time-point measures that may not be reproduced identically on another day. Future research addressing this limitation is discussed in the following section.

6.4 Future Research

Future research should first determine the between-day reproducibility of the static balance classifications and dynamic gait-asymmetry outcomes identified in the present study. Repeating the static and dynamic assessments across multiple days would establish whether participants retain similar classifications and asymmetry values despite day-to-day changes in physical and cognitive condition. This would help distinguish persistent characteristics of postural and gait control from transient variation associated with factors such as sleep, attention, pain, medication effects, fatigue, and recovery status. Recording these factors at each session would also allow their potential contribution to variation in balance and gait performance to be examined.

Prospective studies should then examine whether questionnaire responses, centre-of-pressure outcomes, and gait-asymmetry measures are associated with subsequent falls when assessed within the same cohort. The present study found that many participants classified as Low-risk by STEADI demonstrated elevated sway during objective static testing; however, without prospective monitoring, these participants cannot be considered confirmed false negatives. Recording falls over an extended follow-up period would clarify whether objective static or dynamic impairments provide information about future falls beyond questionnaire-based screening. Future studies should also extend gait assessment beyond straight-line walking. Turning, obstacle negotiation, and dual-task walking place greater demands on gait adaptation, attention, and cognitive-motor integration and may reveal impairments that are not apparent during uncomplicated walking (Maki & McIlroy, 2006; Yogev-Seligmann et al., 2008). These tasks may be particularly informative for determining whether the temporal and loading-related asymmetries identified in the present study become more pronounced when walking demands increase.

Further research should also determine whether the gait-asymmetry variables associated with falls-related vulnerability are responsive to intervention. The present findings identified greater swing-time and average loading-rate asymmetry among participants with poorer quiet-standing control, but the study did not examine whether these outcomes can be modified. Longitudinal training studies could therefore evaluate whether interventions targeting balance, lower-limb strength, gait coordination, and weight acceptance improve static postural control and reduce between-limb asymmetry. Exercise programmes incorporating balance and functional training are known to reduce falls in community-dwelling older adults (Sherrington et al., 2019); however, examining changes in centre-of-pressure and in-shoe gait outcomes would help identify the mechanisms through which training improves performance. Repeated assessment before, during, and after an intervention could also determine whether improvements are retained and whether changes in static balance occur alongside changes in dynamic gait control.

A longer-term research priority should be to evaluate whether questionnaire-based screening and brief objective assessments can be combined to provide a more holistic appraisal of falls-risk. Current international guidance recognises that falls arise from interacting physical, cognitive, clinical, behavioural, and environmental factors rather than from a single impairment (Montero-Odasso et al., 2022). Future studies could test whether including a brief measure of quiet-standing sway and functional gait alongside questionnaire screening improves the identification of older adults with emerging balance impairment. The purpose would not be to replace STEADI or other screening tools, but to determine whether self-reported and performance-based information can be integrated from the beginning of the assessment pathway. Such research may support a practical approach that identifies different sources of falls-related vulnerability and guides older adults towards assessment or intervention appropriate to their individual risk profile.

Chapter Seven: Conclusions

This study examined the relationships among questionnaire-based falls-risk screening, objective static balance assessment, and dynamic gait analysis in community-dwelling older adults. It compared STEADI classification with a falls-risk classification derived from force-plate assessment, evaluated centre-of-pressure characteristics and gait asymmetry between static balance-derived risk groups, and investigated whether gait asymmetry differed following periods of mobility and immobility. It also examined whether selected gait-asymmetry variables contributed to static balance-derived falls-risk classification.

The findings showed that STEADI and the static balance assessment classified participants differently. STEADI identified six participants as High-risk, whereas the static balance assessment identified 21, demonstrating a clear discrepancy between the two approaches. Participants classified as High-risk by the static assessment demonstrated greater centre-of-pressure path length, velocity, frequency, spatial dispersion, and anterior-posterior sway irregularity, indicating clear differences in quiet-standing postural control. They also demonstrated greater asymmetry in selected dynamic outcomes, including ground contact time, swing time, peak impact time, midstance time, and average loading rate. However, group differences were not evident across all dynamic variables, and gait asymmetry did not differ significantly between the Mobile-10MWT and Immobile-10MWT conditions. Mean swing-time and average loading-rate asymmetry jointly contributed to a significant logistic regression model that accounted for 42.9% of the variation in static balance-derived classification according to the Nagelkerke R^2 and correctly classified 75.8% of participants.

Collectively, these findings indicate that questionnaire-based screening, static balance assessment, and dynamic gait analysis provided related but non-equivalent information within this cohort. The discrepancy between STEADI and the static balance-derived classification supports the hypothesis that some participants with elevated postural sway may not be identified as High-risk through questionnaire-based screening alone. The selective gait-asymmetry differences and significant regression model also support the hypothesis that static and dynamic balance were related but not interchangeable. The hypothesis that gait asymmetry would differ between falls-risk groups was supported for selected outcomes, whereas the hypothesis that asymmetry would increase following the

period of immobility was not supported. Under the conditions tested, one minute of sitting did not produce a measurable change in gait asymmetry.

From an applied perspective, the findings support drawing on multiple sources of information when evaluating falls-related vulnerability. The STEADI Stay Independent questionnaire provides information about recognised self-reported falls-risk indicators, whereas portable force-plate and in-shoe assessments provide objective information about quiet-standing postural control and between-limb gait asymmetry. Rather than being viewed as interchangeable, objective performance measures may complement questionnaire-based screening by identifying aspects of balance and gait that are not directly assessed through self-report. Collectively, these findings should be interpreted within the broader context of physical, cognitive, clinical, behavioural, and environmental factors that contribute to falls-related vulnerability. In community and clinical settings, incorporating portable objective balance assessment alongside questionnaire-based screening may help identify older adults who demonstrate postural or gait impairments despite reporting relatively few recognised falls-risk indicators. Combining static postural assessment with selected dynamic gait measures may also provide a broader profile of balance performance than either approach alone. This additional information could help guide decisions about whether further assessment or appropriately targeted balance intervention is warranted, although prospective research is required to determine whether such an approach improves the identification or prevention of subsequent falls.

Future research should examine the between-day reproducibility of static balance classifications and dynamic gait-asymmetry outcomes and determine whether repeated assessment provides a more stable representation of usual performance. Prospective falls monitoring is also required to establish whether the static and dynamic outcomes identified in the present study are associated with subsequent falls rather than assessment-derived classifications alone. More demanding gait tasks, such as turning, obstacle negotiation, dual-task walking, and reactive balance assessment may further clarify how quiet-standing instability and gait asymmetry relate to the balance demands encountered during everyday movement.

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Appendices

Appendix One: Information Sheet

MASSEY UNIVERSITY
TE KUNENGA KI PŪREHUROA
UNIVERSITY OF NEW ZEALAND

School of Sport, Exercise & Nutrition
College of Health
Massey University
Tennent Drive
Palmerston North 4442

Investigating the Relationship Between Falls Risk, Static Balance and Dynamic Balance.

Participant Information Form

Kia ora and thank you for your interest in this study. You are invited to take part in a research project being conducted as part of my master's thesis at Massey University. This study aims to better understand how different types of balance assessments - both static (standing still) and dynamic (walking) - relate to fall risk in older adults. Your participation will contribute to a growing body of research aimed at improving fall prevention strategies in Aotearoa New Zealand.

Why are we doing this? Falls are the leading cause of injury-related hospital admissions among older people in New Zealand. Most fall risk assessments currently used in community settings focus on static balance, such as standing still for a period of time. However, most falls occur while moving. This study is designed to explore whether measuring dynamic balance - how we move and adjust while walking - provides better insight into real-world fall risk. Using specialised equipment, this study will compare both types of balance assessments to see which provides more meaningful information to help reduce falls among older adults.

Who's the research team?

Researcher: Natasha White - Masters student (School of Sport, Exercise and Nutrition, Massey University), [redacted]

Supervisor: Dr. Paul Macdermid (School of Sport, Exercise and Nutrition, Massey University) (06) 9516824; P.W.macdermid@massey.ac.nz

Supervisor: Professor Darryl Cochrane (School of Sport, Exercise and Nutrition, Massey University), (06) 9517532; D.Cochrane@massey.ac.nz

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Who can participate? This study is open to adults aged 65 years and older who live independently in the community. To take part, you must be able to stand and walk without assistance. For safety reasons, people with certain medical conditions or injury such as recent surgery, unmanaged heart problems, or lower limb injuries, may not be able to participate. If you're unsure whether the study is suitable for you, please me to discuss your situation.

What's involved?

If you decide to take part, you will be invited to a single testing session lasting around 15 minutes at Barber Hall in Palmerston North. This is a familiar and accessible venue for older adults. Trained staff will be present throughout the session to support your safety and comfort, and rest breaks will be offered as needed.

The study involves adults aged 65 years and older who live independently.

A brief health questionnaire will be used only to confirm whether you meet the study's inclusion and exclusion criteria. Your responses will not be kept as part of the research data; only whether you are eligible or not will be recorded.

If eligible, you will begin by completing the falls risk questionnaire to identify your level of fall risk.

Then, you will be equipped with force sensor insoles that will be placed in a pair of flat shoes provided by the research team and calibrated based on your body weight. Next, you will complete two types of balance assessments while wearing pressure-sensing insoles inside standardised shoes:

- Static balance will be assessed while you stand on a force plate (a specialised balance mat). You will do four 20-second trials - one practice and three for analysis.
- Dynamic balance will be assessed using two 10-metre walking tests. For the first test, you will complete a one-minute walking warm-up before walking 10 metres at your usual pace. For the second test, you will sit for one minute, then stand up, walk 10 metres, and stop at the end of the walkway.

How long will it take? The full session will take approximately 15 minutes to complete. This includes a brief introduction to the study, time to fill out the falls risk questionnaire, and fitting of the pressure-sensing insoles and standardised footwear. You will then complete a series of balance assessments, both while standing and walking. Rest breaks will be offered as needed.

Benefits of participating? By taking part in this study, you will have the opportunity to receive a personalised summary of your balance results, which may offer useful insights into your movement and stability. In addition, your contribution will support important research aimed at helping health professionals improve how falls are assessed and prevented in older adults. Your involvement may assist in developing better tools and screening strategies that benefit communities across Aotearoa.

How's your data handled and stored?

All information will be handled in line with Massey University's policies on data collection, storage, and security. Digital files will be kept on Massey's Microsoft OneDrive, which is an approved, password-protected, encrypted, and multi-factor authenticated storage system for research data. Any documents containing identifying details, such as consent forms or participant contact information, will be stored separately in a secure folder that only the primary researcher (Natasha White) and, if necessary, supervisors can access. Hard copy data will be kept in a locked filing cabinet in Natasha White's Massey University office. All data - both identifiable and de-identified - will be retained for five years after the completion of the project and then securely destroyed in accordance with Massey University policy. No data will be shared, archived, or used for future research beyond this project. At the point of deletion, digital records will be permanently erased, and physical documents will be cross shredded to ensure confidentiality.

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What are your rights as a participant? Taking part in this research is completely voluntary. You can choose not to answer any questions, ask for clarification at any time, and stop participating whenever you like - without giving a reason. You also have the right to withdraw your data up to two weeks after the session, after which it will not be processed or included in the results. Furthermore, you will receive a brief results sheet outlining your individual assessment results once the data has been analysed. You may also request a summary of the overall study results after the project is completed. This summary will describe the main outcomes of the research and will be emailed to you or made available through Sport Manawatū.

Compensation for Injury

In the unlikely event that you are physically injured as a result of taking part in this study, you should seek medical attention and submit a claim to ACC (Accident Compensation Corporation) as soon as possible. Please note that ACC cover is not automatic - your claim will be reviewed under the Accident Compensation Act 2001. If accepted, ACC will inform you of your entitlements and assist you in accessing them. These may include support such as treatment costs, travel expenses for rehabilitation, compensation for lost income, and possibly a lump sum for permanent injury. Compensation for emotional or psychological distress is only available if it results directly from a physical injury.

If your ACC claim is declined, please contact the researcher immediately. In that case, steps will be taken to ensure you receive compensation equivalent to what you would have received if your ACC claim had been approved.

Ethical Approval

This project has been reviewed and approved by the Massey University Human Ethics Ohu Matatika 1, Application OM1 25/76. If you have any concerns about the conduct of this research, please contact the Chairperson, Massey University Human Ethics Ohu Matatika 1, email humanethics1@massey.ac.nz.

Appendix Two: Consent Form

MASSEY UNIVERSITY
TE KUNENGA KI PŪREHUROA
UNIVERSITY OF NEW ZEALAND

School of Sport, Exercise &
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of Health
Massey University
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Investigating the Relationship Between Falls Risk, Static Balance and Dynamic Balance.

Participant Consent Form

- I understand that this consent form will be securely stored for a minimum of five years and then destroyed in accordance with Massey University policy.
- I have read the Participant Information Sheet and had the purpose and procedures of the study explained to me. Any questions I had have been answered to my satisfaction, and I am aware that I may ask additional questions at any stage and decline to respond to any specific items.
- I understand that I may withdraw from the study at any time. However, I may withdraw my data only up to two weeks after my data collection session, after which point it may not be possible to remove it from the analysis.
- I consent to providing information to the researcher with the understanding that my name or identifying details will not be disclosed without my explicit permission. I acknowledge that the information I provide will be used only for this research project and will not be shared or used for any future studies.
- I agree to take part in this research under the conditions described in the Participant Information Sheet.

Signature: _____ **Date**_____

Full Name (printed) _____

Phone Number_____ **Age** _____ **Date of Birth** _____

Participant code ☐

Appendix Three: Health Screening Questionnaire

MASSEY UNIVERSITY
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School of Sport, Exercise & Nutrition
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Investigating the Relationship Between Falls Risk, Static Balance and Dynamic Balance.

General Health Questionnaire

Participant Name: _____

Section 1: Personal Information

- Age: _____
- Sex: ☐ Male ☐ Female ☐ Other ☐ Prefer not to say

Section 2: Medical History

1. Do you have, or have you ever had, any of the following conditions? (tick all that apply)
    - ☐ heart disease, chest pain, or irregular heartbeat
    - ☐ High or low blood pressure
    - ☐ Stroke or circulation problems
    - ☐ Respiratory issues (e.g., asthma, COPD)
    - ☐ Diabetes (Type I or Type II)
    - ☐ Epilepsy or seizures
    - ☐ Arthritis or joint problems
    - ☐ Neurological conditions (e.g., Parkinson's, MS)
    - ☐ Osteoporosis or bone conditions
    - ☐ Recent surgery or hospitalisation
    - ☐ Other (please specify): _____

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Section 3: Current Health

2. Are you currently taking any medications?
   - ☐ No
   - ☐ Yes (please list): _____
3. Do you currently experience pain, dizziness, or balance problems?
   - ☐ No
   - ☐ Yes (please describe): _____
4. Do you have any injuries or physical limitations that may affect participation?
   - ☐ No
   - ☐ Yes (please specify): _____

Section 4: Lifestyle Factors

5. Do you smoke? ☐ Yes ☐ No
6. Do you drink alcohol? ☐ Yes ☐ No - If yes, how often? _____
7. How often do you usually engage in physical activity?
   - ☐ Never
   - ☐ 1-2 times per week
   - ☐ 3-4 times per week
   - ☐ 5 or more times per week

Section 5: Emergency Information

8. Emergency contact name: _____
   Relationship: _____
   Phone: _____

Section 6: Declaration

I confirm that the information I have provided is true to the best of my knowledge. I understand that this information is collected for the purpose of ensuring my safety during participation.

Signature: _____
Date: _____

Appendix Four: Massey University Human Ethics Committee, Ohu Matatika 1, Application 25/76

TE KUNENGA KI PŪREHUROA | MASSEY UNIVERSITY
UNIVERSITY OF NEW ZEALAND

5/12/2025

Dear: Natasha White

Re: Ethics Application - OM1 25/76 - Investigating the Relationship Between Falls Risk, Static Balance and Dynamic Balance.

Thank you for the above application that was considered by the Massey University Human Ethics Committee:

Ohu Matatika 1 at their meeting held on **Tuesday, 11 November 2025**

On behalf of the Committee I am pleased to advise you that the ethics of your application are approved.

Approval is for three years. If this project has not been completed within three years from the date of this letter, reapproval must be requested.

If the nature, content, location, procedures or personnel of your approved application change, please advise the Secretary of the Committee.

Yours sincerely



Professor Tracy Riley,
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Appendix Five: STEADI Falls-Risk Screening Tool

MASSEY UNIVERSITY
TE KUNENGA KI PŪREHUROA
UNIVERSITY OF NEW ZEALAND

School of Sport, Exercise & Nutrition
College of Health
Massey University
Tennent Drive
Palmerston North 4442

Investigating the Relationship Between Falls Risk, Static Balance and Dynamic Balance.

Fall-Risk Questionnaire

Please complete the following questions to enable us to gain insight into your current falls risk.

1: Please insert your **Name**

_____

2: Please select the ethnic group(s) you identify with

- ○ African
- ○ Asian
- ○ European
- ○ Latin American
- ○ Maori
- ○ Middle Eastern
- ○ New Zealander
- ○ Other

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3: I have fallen in the past year.

- ○ No
- ○ Yes

4: I use or have been advised to use a walking stick or walker to get around safely.

- ○ No
- ○ Yes

5: Sometimes I feel unsteady when I am walking

- ○ No
- ○ Yes

6: I steady myself by holding onto furniture when walking at home.

- ○ No
- ○ Yes

7: I am worried about falling.

- ○ No
- ○ Yes

8: I need to push with my hands to stand up from a chair.

- ○ No
- ○ Yes

9: I have some trouble stepping up onto a curb.

- ○ No
- ○ Yes

10: I often have to rush to the toilet.

- ○ No
- ○ Yes

11: I have lost some feeling in my feet.

- ○ No
- ○ Yes

12: I take medicine that sometimes makes me feel light-headed or more tired than usual.

- ○ No
- ○ Yes

13: I take medicine to help me sleep or improve my mood.

- ○ No
- ○ Yes

14: I often feel sad or depressed.

- ○ No
- ○ Yes