Flinders University

Postural sway in near fallers: effects of fatigue and distraction

By

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ABBREVIATIONS

5TSTST	5 Times Sit to Stand Test
AIHW	Australian Institute of Health and Welfare
AP	Anteroposterior
AUC	Area Under the Curve
AV	Angular Velocity
Base	Baseline testing
BBS	Berg Balance Scale
BESTest	Balance Evaluation Systems Test
BMI	Body Mass Index
CC	Correlation Coefficient
CI	Confidence Interval
CINAHL	Cumulative Index to Nursing and Allied Health Literature
COM	centre of mass
COP	centre of pressure
csv	comma separated value
df	degrees of freedom
DT	distraction
EC	Eyes Closed
EO	Eyes Open
FA	Feet Apart
FMS	Functional Movement Screen
FRT	Functional Reach Test

FT	Feet Together
g	Acceleration of gravity
Gen Hlth	General Health
HR	Heart Rate
Hz	Hertz (unit of frequency; cycles per second)
ICC	intraclass correlation coefficient
ID	Identity
IMU	Inertial Measurement Unit
IQR	interquartile range
ISWT	Incremental Shuttle Walk Test
Lun	Lunge
Max	Maximum
Md	Median
Min	Minimum
ML	Mediolateral
MSLHST	Multiple Single-Leg Hop-Stabilization Test
m/s^2	metres per second squared (unit of acceleration)
n	number in sample
OR	Odds Ratio
PF	prospective faller
Phys Fn	physical function
PICO	Population, Intervention, Comparison, Outcome
PNF	prospective near faller
P-non	prospective non-faller

PRISMA	Preferred Reporting Items for Systematic Review and Meta Analysis
rad/s	radians per second (unit measure of angular velocity)
RMS	Root Mean Square
RNF	retrospective near faller
R-non	retrospective non-faller
ROC	Receiver Operating Characteristic
s	seconds
SE	standard error of the estimate
SF36	36-Item Short Form Survey
Sig	significance
SLS	Single leg stance
SPSS	Statistical Package for the Social Sciences
SSQ5	Speech and Spatial Qualities of Hearing (5 questions)
SW	shuttle walk (from incremental shuttle walk test)
Tan	Tandem steps
TRIPOD	Transparent Reporting for Individual Prognosis Or Diagnosis
TUGT	Timed Up and Go Test
V	Vertical
VM	Vector Magnitude
VM_a	Vector Magnitude of acceleration
WHO	World Health Organisation

ABSTRACT

Near falls are more frequent than, and a precursor to, falls. Near falls are any momentary loss of balance where corrective action prevented a fall. Near falls and falls result from disruptions to postural control. While there is good understanding of postural control from birth through childhood into young adults, and in older adulthood from age 65 years, there is little known about balance in midlife from aged 40 to 65 years. During midlife, people are at risk of covert functional decline that is predominantly related to the physical function factors of muscle mass, strength, and balance.

The control of balance relies on these physical factors and the sensory systems. Balance control is exhibited as changes in postural sway and function. Confounders to balance control include distraction and fatigue. The aims of this research were to 1. understand the contributing factors for near falls, 2. to investigate differences in postural sway between near fallers and non-fallers, under normal, distracted and fatigued conditions, and 3. determine the utility of postural sway to predict near falls. Preparatory work identified clinical balance tests to discriminate near fallers from non-fallers.

Based on previous research with a similar community-based midlife population, near fallers were 2-3 times more likely to fail single leg stance, lunge and five tandem steps forwards than non-fallers. While these clinical tests provided a pass/fail measure of balance, they provided no detail, such as sway direction or magnitude. A scoping review of systematic reviews explored the instrumentation suitable for measuring postural sway in a community setting. A subsequent systematic review found that inertial sensors provided a valid and reliable option to measure postural sway in the community.

Consequently a longitudinal study measured balance outcomes and postural sway using the identified tests in healthy, midlife, community dwelling, non-faller or near faller adults. An initial survey provided self-report measures on demographics, quality of life, hearing, vision and dizziness to answer the study’s first aim. Then balance was tested in single leg stance, lunge and tandem steps, while confounding with distraction or fatigue. Distraction consisted of serial subtraction or categorical naming tasks concurrently with the balance task. Fatigue of the legs occurred following the incremental shuttle walk test. Sway was synchronously measured by a wearable inertial sensor taped to L4. Following the balance testing session, participants completed a daily near falls/falls diary for three months. Outcomes of the diary allocated participants to one of near faller, non-faller, or faller groups.

My unique contributions to knowledge are the explanation of balance using postural sway data in midlife adults; identifying the predictive variables to distinguish near fallers from non-fallers; and determining the differences in postural sway between near fallers and non-fallers.

STRUCTURE OF THESIS

This thesis is divided into nine chapters. The first chapter is an introduction to the thesis. It provides a rationale for undertaking the research and states the research question.

Chapter two summarises the background literature. It presents the context for the problem, setting, equipment and balance activities through the lens of balance control.

Chapter three selects the most appropriate materials and methods to assess balance and postural sway in midlife adults.

Chapter four describes the approach to the study, including the design, population, intervention, outcome measures and statistical analysis plan for the research.

Chapter five displays the demographic results for near falls and explores the differences between retrospective and prospective reports of near falls.

Chapter six provides the postural sway results at baseline and under the distracted and fatigued conditions.

Chapter seven delivers the results of the demographic and sway variables to predict near falls.

Chapter eight discusses the study findings and emphasises the new knowledge in the context of the extant literature.

Chapter nine outlines the clinical implications and future directions.

DECLARATION

I certify that this thesis does not incorporate without acknowledgment any material previously submitted for a degree or diploma in any university; and that to the best of my knowledge and belief it does not contain any material previously published or written by another person except where due reference is made in the text.

Signed   N R Baker

Date........15 July 2022

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"Life is like riding a bicycle. To keep your balance, you must keep moving." Albert Einstein

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PUBLICATIONS ARISING FROM THIS STUDY

Chapter	Publication	Contributors (indicated by initials)
3	Baker, N., Gough, C., & Gordon, S. (2020, Jan). Classification of balance assessment technology: A scoping review of systematic reviews. In A. J. Maeder et al., (Eds.), *Information Technology Based Methods of Health Behaviours*, *(pp. 45-59)*. IOS Press. https://doi.org/10.3233/SHTI200005	This scoping review was conceptualised by NB with support from supervisor SG. The search terms and protocol were written by NB with support from SG. NB conducted the searches, title/abstract screening, and full text selection with assistance from CG. NB conducted data extraction and CG cross-checked data. NB wrote the first draft of the manuscript and all authors contributed to the final version for publication.
3	Baker, N., Gough, C., & Gordon, S. J. (2021, Jul). Inertial sensor reliability and validity for static and dynamic balance in healthy adults: A systematic review. *Sensors, 21*(15). https://doi.org/10.3390/s21155167	This systematic review was conceptualised by NB with support from supervisor SG. The search terms and protocol were written by NB with support from SG. NB initiated the searches, title/abstract screening and full text selection, with assistance from CG. NB conducted data extraction with cross-checking provided by CG. NB wrote the first draft of the manuscript and all authors contributed to the final version for publication.
3	Baker, N., Grimmer, K., & Gordon, S. (2021, Aug). Balance provocation tests identify near-falls in healthy community adults aged 40-75 years: An observational study. *Physiotherapy Theory and Practice*. 1-10. https://doi.org/10.1080/09593985.2021.1983909	This study was designed under supervision of SG as part of a larger project. NB provided input to the ethics application and recruitment. NB collected the data. Initial analysis was conducted by NB with support from SG and KG. Nicky wrote the manuscript with input from KG and SG. All authors had input to the final version.

CHAPTER 1 INTRODUCTION

Near falls are more frequent than falls. People who experience near falls are at higher risk of falling than those who do not (Nagai et al., 2017). Near falls are defined in this thesis as *"any stumble, trip, slip, mis-step or other momentary loss of balance where corrective action prevented a fall"* (Pang et al., 2019, p. 48), whereas falls are defined as *"an event which results in a person coming to rest inadvertently on the ground or floor or other lower level"* (WHO, 2021). Both near falls and falls occur due to a change in balance.

Balance and Postural Sway

Balance can be considered a generic term for attaining, maintaining, or regaining upright stability while sitting, standing, or moving (Pollock et al., 2000 p. 405). Postural sway describes the movement of the body during balance control. Postural sway is defined as a change in position of the centre of mass over its base of support (Horak, 2006). When the base of support is wide, such as standing with feet apart, there is opportunity for the centre of mass to move considerably within the base before reaching the edges. Conversely, when the base of support is narrow, such as standing on one leg, there is less latitude for movement of the centre of mass, before reaching the limit of the base of support. Thus, with a larger base of support there is more stability and less postural sway; conversely, when the base is small, there is less stability and increased postural sway.

Maintenance of the centre of mass over the base of support can be considered in terms of static, dynamic, proactive, and reactive balance. Standing, sitting or holding a position without moving requires postural control. Balance described in a stationary position is termed 'static'. In static balance, the body remains still until there is a force enacted on it. The force will be either intrinsic, meaning it is created within the body, or extrinsic, an external force from the environment. As soon as body movement occurs, the centre of mass also moves, and balance is described in terms of 'dynamic' control. In dynamic balance, the centre of mass moves as the body moves. To maintain an upright position, the centre of mass needs to remain within the base of support, even though the base of support may be constantly changing. Consider walking when the movement forwards is generated by stepping from one foot to the other. As the body weight is on the standing foot, the centre of mass moves over that foot to permit the other leg to swing forward. As the swing leg contacts the ground, the centre of mass moves across to that side to take weight, and so on. As the centre of mass nears the vertical position over the edge of base of support, it reaches the margin of stability, which is defined as the distance from the centre of mass to the limit of the base of support (Hof et al., 2005). Dynamic movement may be intentional, that is, proactive, such as reaching, walking or changing position (van der Kruk et al., 2021). It may also be unintentional, reactive balance, such as stepping into an unseen depression in the ground, or being knocked by a moving object (Horak, 2006). The translation of these concepts to clinical practice is that postural sway increases as the base of support reduces. Sway also increases as the balance activities become more challenging and in populations with known reductions in balance control mechanisms, such as people with neurological conditions (Roman-Liu, 2018). When the centre of mass moves outside the base of support without corrective action, a fall can occur. Falls are costly to the person, the health system and to society.

Falls

The unintentional nature of a fall can result in an injury that causes pain and suffering. Globally, there is a considerable burden from falls in terms of mortality, injury, living with a disability and lost productivity (James et al., 2020). The global death rate from fall incidents (9.2 per 100,000) is second only to motor vehicle injuries (approximately 13 per 100,000) (WHO, 2022). The death rate from falls increases with age, particularly in adults aged 60 years and over (James et al., 2020). In the working population, there is distressing evidence that older workers are more likely to have fatal accidents than younger workers (Bande & López-Mourelo, 2015; Bravo et al., 2022). Industries with a higher accident mortality rate also have a correspondingly higher falls mortality rate (Bravo et al., 2020). In Australia, the fall-related mortality rate is approximately 3.86 per 100,000 and, as with the global trend, the rate increases with age (Wu et al., 2020). Falls cause most of the injury-related deaths in Australia (AIHW, 2022).

The global burden of non-fatal injuries from falls is also significant for living with disability and reduced longevity (James et al., 2020). The primary reasons for disability across all aged groups are lower limb fractures and head injuries, with the former particularly true in adults aged over 60 years (James et al., 2020). Globally, there are over 37 million falls each year that result in hospitalisation (WHO, 2022). This pattern is similar in Australia, where over 40% of all injuries requiring hospitalisation during 2018-19 were caused by falls (AIHW, 2021). Falls-related hospital length of stay is longer than other injury related admissions, impacting both the person with injury and the health system (AIHW, 2022). Falls occur at all ages, with highest prevalence in older adults (WHO, 2021) but midlife adults are also at fall risk (Peeters et al., 2018).

Other than the direct effect of disability from fall injury, there are secondary effects from falls. The pain and suffering can impact mobility, balance and associated confidence. Reduced mobility has knock-on effects of muscle atrophy. The resultant weaker muscles create a further falls risk (Forte et al., 2021). The reduced confidence creates a fear of falling which compounds the cycle of fear, leading to movement avoidance, muscle weakness and reduced balance. This cycle of deterioration can lead to social avoidance or isolation (Merchant et al., 2020). In some cases, the fear avoidance cycle is severe enough to cause residential care admission (Bjerk et al., 2018), or in worst cases, a fatal fall (Wu et al., 2020). One of the precursors to a fall is a near fall (Nagai et al., 2017).

Near Falls

In situations when stability is temporarily lost but balance is regained, the incident is defined as a near fall. Near falls encompass several different concepts. A ‘stumble’ is a broad term to describe a faltering step while ambulant, or “*pragmatically (be) defined as losing balance but regaining it before a fall occurs”* (Wiles et al., 2006, p. 393). Trips are occasions when the foot is caught in, on or by an object or surface, and are often attributed as the most common cause for near falls or falls (Bohrer et al., 2022). After a trip, one of three reactions occurs to prevent falling. The first is to lift the caught leg higher out the way of the object, the second is to lower the leg quickly to gain bipedal support, and the third is to delay lowering (Eveld et al., 2021). ‘Slips’ describe occasions when traction between the foot and the contact surface is unable to be maintained (Safe Work Australia, n.d.). This is commonly due to water in its many forms but may also be due to a surface, such as gravel, that has reduced traction. The friction between foot and surface, necessary to create traction for the body’s forward propulsion, is lost by the pieces of gravel moving on each other (Trkov et al., 2018). A misstep is any unintentional, ill-judged placement of the foot that does not result in a fall (Srygley et al., 2009). All these terms cover interim loss, and subsequent regain, of balance. When people stumble, trip or slip without falling, the occurrence is usually dismissed - no injury or embarrassment occurred, and no intervention was required. While near falls are potentially more common than falls, they are under-reported or disregarded due to lack of subsequent sentinel event and injury (Department of Health, 2020). Given that falls are already underreported (Hoffman et al., 2018), the reliance on recall of near fall events is even less trustworthy. This disregard of near falls misses the opportunity to identify and remediate contributing factors and prevent future falls.

There has been intermittent interest in near fallers over the past 30 years, particularly in older adults (Ryan et al., 1993; Srygley et al., 2009, Nagai et al., 2017; van Dieën et al., 2005), but also in people with hip osteoarthritis (Arnold et al., 2007), Parkinson’s Disease (Iluz et al., 2014) and Multiple Sclerosis (Brandstadter et al., 2020; Fritz et al., 2018). More recent interest has been in the accurate identification of a near fall using sensor technology (Nouredanesh et al., 2021; Pang et al., 2019), and the sensor’s ability to differentiate a near fall from a fall (Aziz et al., 2017; Handelzalts et al., 2020; Wang et al., 2022).

Both near falls and falls occur due to a disruption to postural control. Righting strategies for recovery from disrupted control generally incorporate one of three approaches - an ankle, a hip, or a stepping strategy (Horak, 2006). The ankle recovery strategy generates movement about the ankle and is dependent on strong and flexible calf and dorsiflexor muscles. The hip strategy involves movement at the hip as a response to disrupted balance, and is a bigger movement, permitting increased range of the centre of mass than is possible at the ankle. Finally, the step strategy incorporates a recovery placement of the second foot to increase the base of support.

While there is good understanding of postural control from birth (Boxum et al, 2019) through childhood (García-Soidán et al., 2020), in young adults (Herssens et al., 2018), and in older adulthood (Cavanaugh et al., 2018), there is little known about balance in midlife. As 'midlife' and 'middle age' are often nebulous terms attributed to various age definitions, the interpretation in this study is adults aged 40-64 years as informed by longitudinal studies in Australia (Lee et al., 2005) and Europe (Hajek & König, 2020; Peeters et al., 2018). During midlife, people are at risk of covert functional decline, where incremental changes occur that are not severe enough to require immediate intervention (Brown et al., 2017; Gordon et al., 2020).

Pre-frailty

These physical decrements can contribute to a diagnosis of pre-frailty, where one or two of the five Fried phenotypes are present (Fried et al., 2001). The five phenotypes encompass unintentional weight loss, self-reported exhaustion, weakness, slow walking speed and low physical activity (Fried et al., 2001). The trajectory of change from pre-frail (one or two phenotypes) to frail (three or more phenotypes) has strong psychosocial considerations. However, the trajectory from robust (no phenotypes) to pre-frail contains mutable factors, relating predominantly to physical function (Gordon et al., 2020). These covert changes involve reduced muscle mass, strength, and balance, and occur from age 40 years (Gordon et al., 2020). This provides an opportunity to understand balance in robust and pre-frail midlife and young-older adults (aged 65-74 years) (Lee et al., 2018), who have no obvious balance deficits. The opportunity to identify reversible deficits in a younger age group has implications for healthy ageing (Atallah et al., 2018). In the current climate of COVID-19, functional decline may be accelerated due to the 'deconditioning pandemic', which describes the reduced physical activity and subsequent deconditioning from social restrictions and lockdowns (Gray & Bird, 2021).

Adults in midlife and the younger cohort of older adults are an under-investigated group for balance, falls risk and covert ageing changes (Peeters et al., 2018). Midlife can be a time for detrimental changes in physical function due to onset of chronic conditions (Lai et al., 2019). Further, the hormonal changes due to menopause affect body metabolism in midlife women (Avis et al., 2009; Xu et al., 2020). With menopause, the subsequent reduction in muscle mass and increase in fat mass negatively affect balance (Lee et al., 2019). For midlife men physical activity reduces over time in those who smoke and are obese (Aggio et al., 2019).

There is an established positive association between physical function and physical activity (Dugan et al., 2018). However, major life transitions, such as moving out from home, getting married or becoming a parent, are associated with reduced physical activity (Gropper et al., 2020). Key events in midlife, such as onset of menopause or retirement, have been promoted as sentinel times to increase strength and balance activities to prevent future decline (Skelton & Mavroeidi, 2018). Targeting physical activity behaviours for people in midlife requires both personalised

(Biernat & Piątkowska, 2018) and policy approaches (Brunner et al., 2018). Personal approaches need to consider the socioeconomics, lifestyle and environment, as well as the person's capabilities, opportunities and motivations for the behaviour change (Michie et al., 2011). Policy approaches that enhance physical activity, encourage healthy eating and eliminate smoking will assist in healthy ageing (Brunner et al., 2018). Encouraging healthy living behaviours in midlife and young older adult ages may also impact balance and the risk of future falls.

Therefore, this study targeted near falls in a cohort of midlife and young-older adults. The aim of this research was to investigate balance and postural sway in seemingly healthy, midlife and young-older adults who experience near falls. The overarching research question is "What is the postural sway in near fallers and the effects on their sway with distraction and fatigue?"

Chapter Summary

This chapter provided the context for the study, identified the gap in current knowledge and the subsequent overarching research question. The next chapter considers the roles of balance control and postural sway in near falls.

CHAPTER 2 BALANCE CONTROL

The previous chapter described postural balance as sustaining, attaining, or recovering stability during stance or movement. This chapter describes the body systems that control balance, the determinants that affect balance, and the relationship between balance control and postural sway.

Balance control describes the capacity to adjust the centre of mass over the base of support during everyday activities. This is done by coordinating sensory and muscular systems in response to internally generated movement or to external forces in the environment (Ivanenko & Gurfinkel, 2018). Greater stability requires larger external forces to move the centre of mass outside the base of support (Pollock et al., 2000). The two main dynamic control mechanisms described in the previous chapter are a) anticipating movement and b) reacting to unexpected forces. Anticipatory control incorporates intended changes in position. This includes the muscle preparation to change position, or to respond to an anticipated external force, such as bracing for imminent contact from an exuberant pet dog. Reactive responses occur from unexpected events such as the lurch of a tram as it moves away from the stop. As a response to external perturbation, restorative strategies are to move from the ankle, the hip or to step, depending on the severity of the perturbation and comorbidities in the person (Aftab et al., 2016). The usual restorative balance strategy from a trip, when the foot has caught on an object, is a sharp increase in hip and trunk flexion followed by a stepping strategy (Handelzalts et al., 2020; Qu et al., 2020). Sometimes multiple steps, squatting to lower the centre of mass, or hopping can also be undertaken to restore balance (Cheng & Yeh, 2015). These movements require a good range of joint motion in the back, hips, knees and ankles, as well as strong, efficient muscle action for a quick response.

Intrinsic Control

Balance is controlled by coordination of the intrinsic sensory, biomechanical, and cognitive systems that work together to maintain the upright position (Horak, 2006). The major sensory systems controlling balance are vision, the vestibular system and proprioception (Bronstein, 2016). These are complemented by hearing (Ojie & Saatchi, 2021), foot sensation (Chimera & Larson, 2020) and coordination (Liu et al., 2020). None of the systems work in isolation but align with each other to produce subtle changes in muscle contraction, and shifts in the centre of mass, to keep the body stable.

Vision

Overall, the two main functions of vision in balance are to provide feedback to the brain on the head’s position relative to its surrounds, and to identify safe terrain and potential obstacles while moving (Hollands et al., 2017). Balance relies on both central and peripheral vision. In central vision, the visual field surveys the ground and surrounds in front of the body. The reliance on the visual field increases with age (Kunimune & Okada, 2017). Visual acuity, or the sharpness of an image, scans the surrounds for potential problems (Bronstein, 2016). Visual acuity has a direct relationship with postural sway, with worse acuity related to increased postural sway (Hunter et al., 2020). Depth perception provides feedback on the proximity of obstacles and hazards and their relationship to the surrounds. Depth perception provides time to prepare or adjust the centre of mass for avoidance i.e., anticipatory postural control. Contrast sensitivity is the ability to see an object against its surrounds. Extremes in light conditions - very bright or low, dim lighting - reduce the contrast sensitivity and increase likelihood of not seeing a hazard that may present a tripping or slipping risk. Central vision also provides feedback regarding the body's position relative to the horizon, reinforcing the upright perpendicular to the earth (Bronstein, 2016). Poor vision can require corrective lenses, such as wearing glasses. If the glasses are multifocal they provide two distinct distances for focus - one close and one distant - which can impact walking balance, particularly step length and foot placement (Bist et al., 2021). The latter is particularly important for walking on uneven ground or moving up and down steps or stairs.

Peripheral vision is also important for balance. Peripheral vision provides feedback about the lateral surrounds while the person is moving (Kim & Park, 2016). Similarly, the peripheral vision system contributes to maintaining the body's stable upright position while the person is static but the environment is moving e.g., standing on a moving tram (Horiuchi et al., 2021).

When the eyes are closed, the body reacts to anticipated problems, moving more over the base of support. Sway therefore increases with eyes closed (Nardone & Turcato, 2018). Balance in people with blindness and partial sight is compensated by increased reliance on the vestibular and proprioception systems (Moghadas Tabrizi et al., 2022).

Vestibular System

The second major sensory system to control balance is the vestibular system, situated in the inner ear. The vestibulocochlear nerve, the eighth Cranial Nerve (CN VIII), provides sensory functions for the vestibular as well as hearing system. The CN VIII provides feedback to the brain about the head's position, its movement, and its orientation relative to gravity (Bronstein, 2016). Information on the angle and speed of the head's movement coordinates with the visual system to assist with the body's postural alignment to maintain an upright position.

The vestibular system has three components. Three semicircular canals in the vestibular system are orientated to the three planes of the human body. The canals are filled with fluid. When the head rotates, the fluid washes over the small hair-like protrusions lining the canals to provide the brain with information on head angular acceleration. The other two components of the vestibular system, the utricle and saccule, provide feedback to the brain on linear acceleration for movement forwards, backwards and sideways. These feedback systems combine with sensory input from joint receptors and movement receptors throughout the body to maintain the body's upright position (Bronstein, 2016).

One of the symptoms associated with the vestibular system is dizziness. Dizziness is a subjective sensation, a symptom triggered by a number of potential central, peripheral or functional reasons. These may include postural hypotension, vestibular disorders or psychological causes (Menant et al., 2020). Dizziness affects up to 35% of the adult population (Jahn, 2019) and is considered a major contributor to fear of falling (Song & Lee, 2020). It negatively affects functional activity (Menant et al., 2020) and health-related quality of life (Lindell et al., 2021). Dizziness impacts balance by the sensation that the body or the room is spinning. Dizziness may also have a secondary effect on balance by restricting physical activity (Morimoto et al., 2019).

Hearing

Hearing influences balance by the person responding to auditory cues in the environment (Berge et al., 2019). People with hearing loss, having an associated decrease in sound from the immediate surrounds, have greater difficulty maintaining postural balance than people with normal hearing (Horowitz et al., 2020). The long term effects of hearing impairment increase the risk of falls (Ogliari et al., 2021b). However, there is conflicting information on the use of hearing aids in relation to balance. Some recent findings suggest that balance is assisted by hearing impaired people wearing hearing aids (Ninomiya et al., 2021), while others suggest hearing aids provide no benefit to the risk of falls (Riska et al., 2021).

Standalone hearing issues are less likely to negatively affect postural control than when there is simultaneous demand on the other sensory and cognitive systems (Carpenter & Campos, 2020). Balance control is reduced if both vision and hearing deficits occur concurrently (Ogliari et al., 2021b). In midlife women, hearing and vision impairment, either independently or concurrently, reduce balance control and increase falls risk. However, in midlife men, only concurrent hearing and vision impairment produce an increased risk of future falls (Ogliari et al., 2021b).

Proprioception

The third major controller of postural balance is proprioception. This is the sensory system that provides feedback to the brain on static joint position and dynamic joint movement (Chiba et al., 2016). In particular, the proprioceptive feedback in the ankles is influential in maintaining balance and is directly related to the sensation on the sole of the foot (Yang et al., 2022).

Touch

The sensation of touch, or tactile sensation, provides feedback on the weightbearing surface, usually the soles of the feet in contact with the ground. The mechanoreceptors in the skin provide feedback on pressure and weight-bearing information to modulate balance (Viseux, 2020). As the body moves during postural sway the pressure distribution also moves and provides the brain with information on the changes in surface contact and movement of the skin. This feedback loop contributes to keeping the body vertical (Viseux, 2020). When foot sensation is reduced, such as with neurological conditions or diabetes, the control of balance is simultaneously reduced (Viseux 2020).

The integration of the multiple sensory systems is required for good balance control, so that when one system is unable to work at full capacity, the other systems take more of the burden to keep the person upright and stable. The sensory systems coordinate with the biomechanical systems for maintaining postural control.

Biomechanical System

The biomechanical system incorporates the muscles, joints and coordination in the body. The biomechanics of muscle strength, power and flexibility are necessary for balance control. Muscle strength is the ability to overcome resistance using force, such as holding the body upright against gravity, standing up from a sitting position, or lifting the leg high enough to step over an obstacle. Muscle power is the strength of muscle in relation to time, or the speed of the muscular contraction. An example of balance-related power would be the ability to contract the muscles strongly and quickly enough to prevent a fall. For balance control, lower limb strength and power are particularly important, especially in the anti-gravity muscle groups of quadriceps, gluteals, hamstrings and calf muscles (Sherrington et al., 2019). However, muscle strength and power reduce with physical inactivity (Elam et al., 2021; Trombetti et al., 2016).

In women, low muscle mass in early adulthood, as well as the loss of mass that occurs from early adulthood to midlife, is related to poor functional balance in later years (Wu et al., 2017). Sarcopenia, the loss of skeletal muscle mass, occurs from midlife and has a corresponding negative effect on balance (Kim et al., 2020). When physical activity or exercise are not maintained from early adulthood, the loss of muscle mass and aerobic fitness affects people in midlife and older age (Edholm et al., 2021). All these findings were supported by Okabe et al., (2021), who identified that loss of muscle mass directly affected muscle strength, balance and functional walking in midlife men and women.

Core strength, the ability to stabilise the centre of the body against movement in the limbs, also contributes to standing and dynamic balance. Of the core muscles (internal and external obliques, multifidus, erector spinae, rectus abdominis and transversus abdominis), the most important for core stability are the internal obliques (Oliva-Lozano & Muyor, 2020). The need for strong core stability increases as the base of support narrows, or the standing surface becomes less stable (Calatayud et al., 2015). Importantly, core stability training can improve balance outcomes in young (Szafraniec et al., 2018) and older adults (Ponde et al., 2021; Sannicandro, 2020).

Other than muscle strength, adequate joint and soft tissue flexibility is necessary to accommodate the range of movements required to change position in dynamic control of posture. Subsequently, joint stiffness (Cenciarini et al., 2010) or tightness in the hamstrings (Fereydounnia et al., 2022) and calf (Costa et al., 2009) are detrimental to postural control.

Age-related Changes

Balance can be affected by biological changes and diseases that onset with ageing (Wu et al., 2021). In midlife adults, chronic conditions such as diabetes and arthritis become more prevalent (Lai et al., 2019). Diabetes is associated with decrements in lower limb muscle strength and sensation (Kraiwong et al., 2019), as well as reductions in vision and hearing, all of which impact balance control (Fasching, 2019). Osteoarthritis is a degenerative condition of the joints with symptoms of pain, muscle weakness, stiffness and reduced function (Assar et al., 2020). These symptoms affect balance control (Lawson et al., 2015) and compound the reductions in muscle strength and power that occur with increasing age (Elam et al., 2021). Other chronic conditions such as cardiovascular disease and Chronic Obstructive Pulmonary Disease can limit physical activity which leads to muscle mass and strength reduction. Inefficient circulation negatively affects the muscles and cognitive function, both of which contribute to reduced balance (Park et al., 2020). Chronic disease conditions are positively correlated with increased likelihood of falls (Paliwal et al., 2017). Chronic disease and multiple co-morbidities often create the need for polypharmacy i.e., multiple medications, and this is an independent risk factor for reduced balance and higher falls risk (Bareis et al., 2018). There are perhaps less obvious but no less important connections between likelihood of falls and bladder incontinence due to increased urgency and increased frequency (Chiarelli et al., 2009). Women post-menopause are more likely to be affected by chronic conditions due to the reduced production of oestrogen (Xu et al., 2020). This is demonstrated by the increased incidence of falls in women aged 45-50 years (19.1%) compared to aged 40-45 years (8.7%) (Peeters et al., 2018). Different ages have different risks associated with fall likelihood, e.g., drinking high levels of alcohol affects the 60th decade more than others (White et al., 2018). Although there is a higher body mass index (BMI) cut-off for overweight in adults aged over 65 years (Kiskaç et al., 2022), obesity and overweight are contributing factors for many chronic conditions, and also increase the risk of falls (Ogliari et al., 2021a). In people who are obese, the proportion of fat to muscle is increased. This shifts the centre of mass anteriorly, affecting standing position and postural control. The subsequent reduced stability and increased sway increases falls risk (Pagnotti et al., 2020). Being underweight is also a falls risk (Ogliari et al., 2021a).

Older adults display a wider stance during gait than younger adults, thereby increasing the base of support (Osoba et al., 2019). Muscle mass often reduces with age, affecting muscle strength and power (Pasco et al., 2020). Physical activity and function reduce proportionally to muscle strength and power loss (Zymbal et al., 2022) and the cumulative effects result in a reduction in balance

(Duck et al., 2019). The changes in musculature and strength affect posture which can become increasingly flexed, throwing the centre of mass forward (McDaniels-Davidson et al., 2018).

Further to the musculoskeletal effects of ageing, our sensory systems also undergo changes over time. Vision impairment incidence increases with age (Flaxman et al., 2017). Visual acuity, depth perception, peripheral vision, resistance to glare and amplitude accommodation each deteriorate over time (Saftari & Kwon, 2018). Sensory processing changes related to ageing include reduced vibration sense and proprioception (Chen & Qu, 2019), touch sensation (Yümin et al., 2016) as well as sensory processing (Chen et al., 2019). As proprioception deteriorates over time, there is associated reliance on vision. Age-related change in the vestibular system reduces the numbers of vestibular cells and neurons, leading to reduced balance and higher risk of falls (Iwasaki & Yamasoba, 2015).

The age-related degradation in all sensory systems may also increase cognitive load and divert attention from the balance activity (Chiba et al., 2016). Divided attention, or distraction, has a negative effect on maintaining balance (Lau et al., 2021). Distractions may be predictable, such as walking while talking, or unpredictable, such as a moving in a noisy or unfamiliar environment. Divided attention demands an increase in cognitive load (Lubetzky et al., 2021). Cognitively, the ability to respond rapidly to stimuli reduces with age, and similar deterioration occurs with processing new information (Thillainadesan et al., 2020). Reduced response time and executive processing both impact the reactive aspects of balance (Solis-Escalante et al., 2019). In older adults, the ability to shift between automatic and focussed attention is better in people with good baseline balance (Kal et al., 2022). Purposeful distraction for balance testing and training employs dual tasks i.e., concurrent cognitive tasks are undertaken during balance testing. Dual tasks produce a reduction in quality of both the cognitive and motor tasks, called ‘dual task cost’. This is caused by overloading the attention or sequence processing ability of the brain (Kiss et al., 2018). During dual task balance, the automatic control of posture increases while the attention is diverted to the cognitive activity (Saint-Amant et al., 2020). Therefore, dual tasks have a detrimental effect on functional activities, such as standing, walking, turning, or stepping, while the focus is on the cognitive process (Bridenbaugh & Kressig, 2015). Dual task balance deteriorates as age increases (Brustio et al., 2017) and dual task testing is more sensitive than balance tasks alone in identifying fall risk in older adults (Ghai et al., 2017).

Aside from distraction affecting balance control, both mood and fatigue are also influencing factors on postural control. Mood, or affect, particularly depression, has a negative association with postural balance in younger men (Stuart et al., 2015), younger women (Williams et al., 2015) and both sexes in the elderly (Casteran et al., 2016). Anxiety, particularly fear of falling, reduces confidence in mobilising (Litwin et al., 2018) thereby having a negative effect on frequency and duration of transfers and walking, leading to an overall reduction in physical activity (Yu Shiu et al., 2022). The cyclical process of reduced activity leading to reduced muscle mass and strength compounds the fear of falling (Litwin et al., 2018). This has a knock-on effect on reducing community ambulation, increasing the risk of social isolation and depression (Zhang et al., 2021). Affect is influenced by sleep quality (Triantafillou et al., 2019). Poor sleep quality contributes to poor postural control in young (Batuk et al., 2020; Tanwar et al., 2021), midlife women (Hita-Contreras et al., 2018) and older community dwelling adults (Takada et al., 2018).

A secondary effect of sleep deprivation is fatigue. Fatigue is a well-accepted non-pathological factor impacting postural sway (Paillard, 2012) and contributing to near falls and falls (Qu et al., 2020). Fatigue can be broadly divided into physical and mental. Physical muscle fatigue results in decreased motor firing and subsequent reduced ability to contract efficiently (Enoka & Duchateau 2016). This leads to impaired muscle coordination and contractility, resulting in reduced motor control and joint proprioception (Abd-Elfattah et al., 2015). The resultant loss of neuromuscular control (Larson & Brown, 2018) alters balance and walking (Kao et al., 2018). Physical fatigue induced by aerobic exercise (Güler et al., 2020), anaerobic exercise (Johnston et al., 2017) and high-intensity intermittent exercise (Whyte et al., 2015) each negatively affect postural control. Trunk fatigue also reduces postural control (Ghamkhar & Kahlaee, 2019). Most of these affect reactive balance, the ability to react to a balance perturbation, whereas neither physical nor cognitive fatigue were shown to affect anticipatory postural adjustments (Schouppe et al., 2019). The effect of calf muscle fatigue on postural control was countered by ankle flexion in young adults, but increased hip, knee and ankle flexion in older adults (Boyas et al., 2019) indicating a bigger movement and more muscle recruitment in older age.

Mental fatigue produces negative changes in emotional status, engagement and/or ability to undertake tasks (Brahms et al., 2022). The effects of mental fatigue on postural control are similar to those of physical fatigue, whereby sway area and speed increases (Morris & Christie, 2020) and the automatic responses for postural control are reduced (Hachard et al., 2020).

Extrinsic Factors Affecting Balance Control

Automatic responses to postural control are challenged by the environment and other extrinsic influences. The less predictable the environment, the greater the risk of falls and requirement for increased balance control (Lee, 2021). The external environment incorporates any surrounds including but not limited to the ground surface, obstacles or obstructions, the available light, ambient temperature, inclement weather and surrounding traffic, for example.

Ground surfaces may be uneven, offering inconsistent contact for the feet, leading to more demands on proprioceptive responses (Riva et al., 2019). Walking on soft sand at the beach provides a less stable surface compared to firm sand. The ground may be slippery, providing less traction for the contact foot. Inclement weather can create other slippery situations such as ice or snow, providing less traction for ambulation. Slippery conditions inside the house may be due to water droplets on tiles in the wet areas of the home.

Obstacles may be inside, such as cluttered furniture, or outside, such as tree roots lifting pavements. Similarly, obstacles may be in the form of other bodies moving in the vicinity, such as other people, animals, vehicles etc. If these are not seen in time to respond and allow balance adjustments, they can cause a tripping or bumping hazard. Further, with extreme low or bright light, visual acuity is hindered by lowered contrast sensitivity and depth perception, making balance control more difficult. Sometimes the contact surface is stable but what lies underneath may not be, for example standing on a paddle board or moving around on a boat.

Other extrinsic factors include ill-fitting clothing or footwear that may be too long or too large, providing little support (Reutimann et al., 2022). Wearing high heels reduces the base of support and shifts the centre of mass forwards, reducing dynamic stability (Chien et al., 2014). While multiple intrinsic and extrinsic factors after balance, this thesis focuses on the intrinsic factors of distraction, fatigue and postural sway.

Chapter Summary

This chapter described the intrinsic control mechanisms and extrinsic influences on postural control and sway. The next chapter explores the most appropriate materials and methods to assess balance and postural sway in midlife and older adult near fallers.

CHAPTER 3 BALANCE AND POSTURAL SWAY MEASURES

The previous chapter identified that balance control is essential for maintaining an upright position, moving successfully, and responding to disruptions from the surrounding environment. It defined postural sway as the changing position of the centre of mass over the base of support. The aim of this chapter was to select the most appropriate materials and methods to assess balance and postural sway in midlife and older adult near fallers. The aim was therefore identified as three objectives:

1. to identify the most appropriate instrumentation for testing postural sway in near fallers
2. to investigate the validity and reliability of the chosen instrumentation for measuring static and dynamic balance
3. to establish the most suitable clinical tests for discriminating near fallers from non-fallers in healthy adults.

Instrumentation Selection

To meet the first objective, the multiple instrumentation technologies that measure postural control and human balance were scoped. Scoping reviews identify the breadth of literature in a particular topic (Munn et al., 2018), therefore a scoping review of systematic reviews was undertaken (Baker et al, 2020; full manuscript attached as Appendix A). The aim of the scoping review was to identify the best option for instrumented postural sway assessment in community dwelling, healthy adults. Medical subject heading (MeSH) terms and keywords were generated for four concepts: postural balance; instrumentation; reproducibility of results; and systematic reviews and endorsed by a research librarian. To gather the most recent information on the current instrumentation, the scope was limited to the previous five years. The databases Medline, Embase, CINAHL, SCOPus and PubMed were searched from 2013 to August 2019 using the Preferred Reporting Items for Systematic Review and Meta-Analyses (PRISMA) extension for scoping reviews (Tricco et al., 2018).

Systematic reviews were included if they had reported instrumented methods of assessing healthy adults' static and dynamic postural balance and were written in English. Exclusions applied to alternate types of balance (e.g., visual balance in artwork), non-human stability or equilibrium (e.g., stability of turbines), fall detection methodology, non-instrumented assessment (e.g., only clinical testing), or intervention rather than assessment. The titles, abstracts and full texts were screened by two independent reviewers, with a third reviewer available for mediation in case of disagreement (Baker et al., 2020, Appendix A).

The search generated 792 systematic reviews, which were filtered according to the PRISMA statement (Moher et al., 2009) (see Figure 1 in Appendix A). Full text screening culminated in 44 papers being included in the review.

Categorisation of devices was informed by the classification system nominated by Chaccour et al., 2017. This provided three groupings: fixed technology that the participant stepped onto or along, such as force plates; wearable technology where sensors were incorporated into clothing, footwear or items worn on or by the participant; and fusion technology which incorporated a combination of wearable and fixed components (Chaccour et al., 2017). A summary table of the advantages and disadvantages found in the scoping review are outlined below (Table 3.1 Instrumentation advantages and disadvantages).

Table 3.1 Instrumentation advantages and disadvantages

Classification system	**Examples**	**Advantages**	**Disadvantages**
Fixed	Gait labs, force plates, fixed video cameras. Posturography; stabilometry	Gold standard benchmark, comprehensive, precise.	Expensive; limited by space; unable to use in community; variability of protocols
Wearable	Inertial measurement units, insoles, clothes	Affordable, discreet, adaptable.	Variability of protocols and body positioning
Fusion	Nintendo Wii, Xbox Kinect, IREX	Interactive, engaging, good for rehabilitation	Variability of protocols

The main advantage of the gold standard accuracy of the fixed instrumentation, such as force plates, was countered by the high costs for the equipment, the proficiency required for image interpretation and venue-specific availability. These disadvantages reduced the attractiveness for using in community-based research, hence they were not further considered.

Fusion systems could be fixed or portable. Fixed fusion systems included gold standard gait labs which assessed balance via instrumented walkways while simultaneously capturing the whole-body movements with video cameras. These were likewise unsuitable for community-based research due to the lack of portability. The portable fusion options incorporated gaming systems such as Xbox Kinect or Nintendo Wii. Gaming for balance has been used increasingly for rehabilitation and reablement; however, there were no standard dosages or protocols, and the emphasis for these systems has been for treatment rather than assessment. This combination also made fusion systems less attractive for community-based assessment of near fallers.

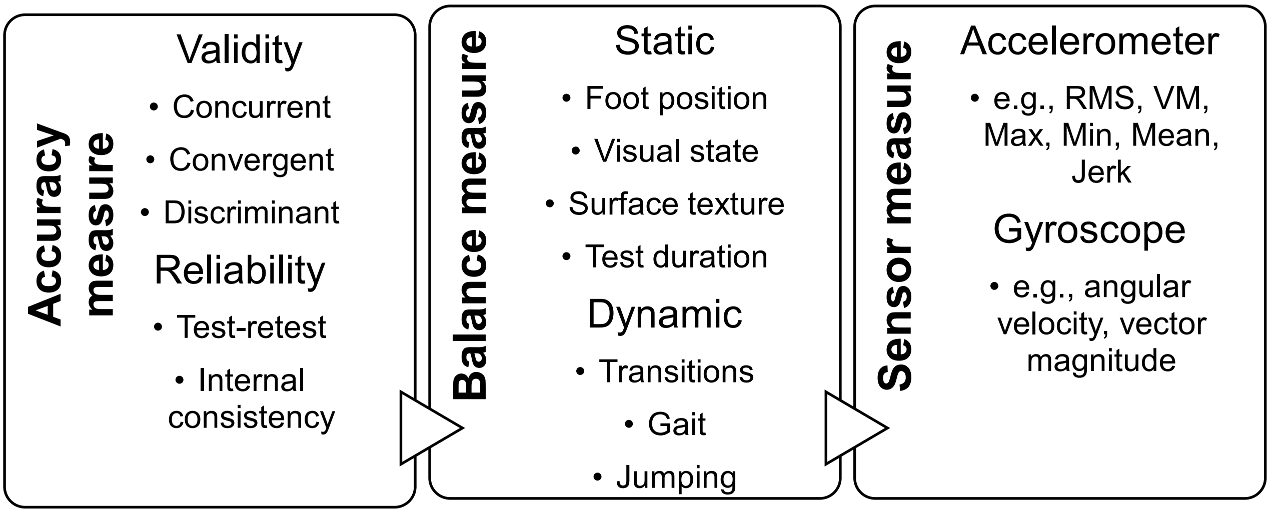
Wearable systems included pressure sensors in shoes and inertial measurement units (IMUs) incorporated into clothing or other wearable devices. The shoe sensors measured pressure distribution through the sole of the foot. The IMUs measured movement across three axes and could be integrated in mobile phones or fitness tracking devices. Disadvantages with the wearable systems included no consistent methods for sampling frequency, body placement position, data extraction or interpretation. However, wearable systems showed several advantages, such as low cost, unobtrusive positioning on the body, and flexibility with accelerometer or gyroscope outputs. Another important advantage was the portability into the community, ensuring ecological validity. The IMUs were able to detect differences in static and dynamic balance in younger adults (Ma et al., 2016), older adults (Pang et al., 2019) and frail elderly (Gordt et al., 2017). The IMUs were also able to detect near falls (Pang et al., 2019) although these were in the laboratory setting. Altogether, the use of an IMU for community-based balance testing was considered an appropriate instrumentation choice for this study. However, the identification of IMUs as a suitable tool did not provide evidence on their reliability and validity to measure postural sway.

Validity and Reliability of Measuring Tool

Therefore, a systematic review of the literature was conducted to investigate the validity and reliability of IMUs for measuring postural sway during static and dynamic balance (Baker et al., 2021b; attached in full manuscript as Appendix B). The study search strategy and selection of manuscripts were conducted according to the PRISMA guidelines (Moher et al., 2009).

Keywords and MeSH terms aligned to Population, Intervention, Comparison, Outcome (PICO) conventions: population of healthy adults; intervention of measuring postural balance by wearable inertial sensor; control as another technological or clinical measure of balance; and outcome being a measure of validity, reliability or accuracy. Five databases were searched - Scopus, Web of Science, PubMed, Embase, CINAHL and Medline. Publication dates before 2010 were excluded due to sensor development since. Other exclusion criteria were applied for non-human or child subject matter, investigation not relating to human balance, pressure or smartphones sensors, and only static or dynamic balance rather than both. Two research staff conducted each stage of screening titles, abstracts and full texts using Covidence systematic review software. A third reviewer was available for arbitration but was not required. The full methodology is provided in the manuscript, Appendix B. Data extraction was cross-checked for correctness. Extracted data were pooled according to the following diagram (Figure 3.1 Data pooling strategy), starting with the validity or reliability component, then the balance static or dynamic activity, and finally the accelerometry or gyroscope measure.

Figure 3.1 Data pooling strategy



Key: Max maximum; Min minimum; RMS root mean square; VM vector magnitude

There were 19 studies included in the review. Healthy adults were described as the sole population in three studies (Heebner et al., 2015; Leiros-Rodriguez et al., 2016; Martinez-Mendez et al., 2011) and as comparators or controls in the remainder. Fallers were the population of interest in four studies (Greene et al., 2012; Liu et al., 2012; Rivolta et al., 2019; Tang et al., 2019), but no studies investigated near fallers.

Balance test activities varied greatly across studies. Static balance incorporated various foot placement positions, such as feet together or apart, single leg stance or tandem stance. The time duration for holding the static position differed across the studies. Further, the sensory input varied according to eyes open or closed and standing on firm or soft surfaces. Dynamic balance activities incorporated gait, functional transfers or changing direction while moving. The outcome measures were similarly variable, using a range of clinical outcomes, e.g., stride length and gait speed. This challenged data pooling for meta-analysis, as demonstrated in the manuscript (Appendix B).

Similarly, there were diverse sensors used, and various methods to capture and extract the sensor data. Each sensor incorporated an accelerometer and 12 of the 19 also contained gyroscopes (See Table 3, Appendix B). The sensor was placed on various parts of the body and the number of sensors ranged from one to six on each person, further demonstrating variability between studies. Data extracted from the sensors were generally analysed through vector magnitude acceleration, root mean square acceleration, and angular velocity for gyroscopic measures (Appendix B).

The sensor was compared to force plates or gait labs as the benchmark to investigate the concurrent validity. For static balance, sensors showed moderately strong validity for anteroposterior sway (r = 0.71) (Martinez-Mendez et al., 2011), and mediolateral sway (r = 0.58 - 0.84) (Heebner et al., 2015; Green et al., 2012; Rivolta et al., 2019). In dynamic balance, good to excellent correlations were demonstrated for gait velocity (Intraclass Correlation Coefficient (ICC) 0.90-0.94), step length (ICC 0.68-0.89) and step time (ICC 0.68 - 0.92) (Heebner et al., 2015; Dalton et al., 2013; O'Brien et al., 2019). These results demonstrated that the inertial sensors were a viable and valid option when compared to the gold standard instrumentation.

Convergent validity assessed the inertial sensor sway outcomes against the clinical test results. Investigations using sensors demonstrated subtle differences between diagnostic groups and healthy controls (de Vos et al., 2020; Spain et al., 2012), and between fallers and non-fallers (Liu et al., 2012; Rivolta et al., 2019) when clinical measures could not. This was a useful finding for application to the unknown near faller population.

Confidence in the sensors to discriminate between groups was encouraging. Fallers were correctly classified from non-fallers with 72-89% accuracy (Greene et al., 2012; Rivolta et al., 2019), and healthy controls from the diagnostic groups with 68-96% accuracy (Spain et al., 2012). However, while accuracy was acceptable for identifying fallers from non-fallers and healthy controls from populations with a medical diagnosis, there was no investigation to compare near fallers to non-fallers, nor subgroups of a healthy population (Appendix B), reinforcing the gap to be investigated in this study.

The test-retest reliability of the inertial sensors showed good results. Results indicated that the IMUs were reliable for measuring dynamic balance (ICC 0.70-0.94) as well as static balance (ICC 0.57-0.79) (Craig et al., 2017; Hasegawa et al., 2019; Heebner et al., 2015). Overall, the systematic review results indicated that inertial sensors were reliable and valid for measuring postural sway in healthy adults in the community. Further, inertial sensors have been shown to identify functional daily activities such as walking up and down stairs and changing posture from sitting to standing to lying (Yen et al., 2020). The sensors' ability to identify subtle sway changes provided encouragement that the differences in sway of near fallers compared to non-fallers might be detectable. This led to the third objective, which was to identify suitable balance tests to distinguish near fallers from non-fallers and fallers.

Clinical Measures

By definition, near fallers have experienced some sort of trip, slip or misstep, but have not fallen. Therefore, in studies investigating only fallers and non-fallers, they would be grouped as non-fallers. In adult populations, falls risk can be the primary aim of balance testing. This is particularly true across different diagnostic groups, and especially for people with neurological conditions. For example, in patients with stroke, clinical tools such as the Berg Balance Scale (BBS) and the Timed Up and Go Test (TUGT) assess static and dynamic balance as a measure of functional recovery and a marker for falls risk (Arienti et al., 2019; Bhatt et al., 2019). In progressive neurological disorders such as Parkinson's Disease, the associated rigidity, bradykinesia and tremor slow down reactions to changes in equilibrium, creating a falls risk. Balance in these populations can be measured by functional activities such as the Functional Reach Test (FRT), or by assessing gait quality with tests such as the Functional Gait Assessment (Osborne et al., 2022). The rehabilitation and recovery required from a physical insult such as hip fracture include measures of stability and function, for example with the Balance Evaluation Systems Test (BESTest) (Miyata et al., 2021). For these populations with a medical condition or diagnosis, the aim of balance assessments is to identify discrepancies so that rehabilitation can be directed at mitigating deficits and restoring functional balance.

However, falls risk is only one component of balance testing. Balance screening and testing in midlife adults is under investigated, as previously outlined. Given the covert nature and gradual changes affecting function in this age group (Brown et al., 2017), balance screening could also be used as a component measure of fitness (de la Motte et al., 2019; Silva et al., 2020), strength (Dixon et al., 2018) or athletic ability (Brachman et al., 2017). For this type of balance testing, the dynamic and proactive balance aspects are assessed in terms of agility. For dynamic and proactive balance, assessments such as the Star Excursion Balance Test and Y-balance test measure the multi-directional reach away from the base of support (Powden et al., 2019). Some balance tests have a higher degree of plyometrics, which incorporate rapid eccentric and concentric muscle actions to produce explosive movements for changes in direction. These are necessary for sudden accelerations or quick changes in direction, jumping or leaping, therefore are useful to assess balance in sportspersons or emergency responders. The Multiple Single-Leg Hop-Stabilization Test (MSLHST) investigates stability in take-off and landing when hopping forwards and diagonally (Sawle et al., 2017). Similarly, the Functional Movement Screen (FMS) assesses mobility and strength during seven functional movements in upper body, lower body and trunk (Scudamore et al., 2019).

In healthy adult populations a functional approach is often preferred, due to postural control being a component of daily living. Balance tests have been used for each of the components of falls risk screening, as well as measures of fitness and of function. While gait speed can be used to assess falls risk (Greene et al., 2019), this thesis focuses on balance. The BBS, TUGT and five times sit-to-stand (5TSTS) are commonly used and concurrently provide a measure of falls risk in adults aged 65 years and older (Lusardi et al., 2017); however, they have not been validated in midlife adults. In midlife, longitudinal studies have investigated falls risk associated with the physiological aspects of ageing (Gale et al., 2016; Hajek & König, 2020; White et al., 2018). While these longitudinal studies rely on self-reports of falls, there is no specific battery of tests for this age group. Recent clinical tests reported for this population include single leg stance with eyes closed for 30 seconds (Okabe et al., 2021), the TUGT, FRT, lateral reach test, step test (Wang et al.,

2021) and the sternal push test (Segaux et al., 2021). For testing balance in a community setting, the chosen balance tests need to be safe, accommodate various levels of mobility or fitness and be easily conducted without complex equipment (Pardasaney et al., 2012).

While clinical tests indicate the prediction for falls, there are no tests that provide an indication for near falls. Despite the volume of research on clinical balance testing in various populations and age groups, there was no standard way to assess midlife adults, nor near falls. Therefore, previous investigations of community healthy ageing by this research team provided deidentified data for analysis on balance test outcomes in fallers, near fallers and non-fallers. These results were analysed to identify a new set of balance tests to distinguish near fallers from non-fallers and fallers (Baker et al., 2021b; manuscript provided as Appendix C). In 2017-18, a healthy ageing project assessed covert signs of ageing in adults aged 40-75 years living in the community. This observational study evaluated social, psychological, and physical aspects of healthy ageing. The full details of the study methodology are already published (Gordon et al., 2019). Participants completed surveys and undertook face-to-face testing. Included in the testing were balance, strength and fitness tests (Gordon et al., 2019). At the end of testing, participants were supplied with a report, comparing their results to the standardised norms. As well as commending the parameters within normal ranges, the report also provided links to resources or free services to address any deficits (Gordon et al., 2019).

As part of the survey, participants were asked if they had experienced any falls or near falls in the previous six months. Participant responses of ‘yes’ or ‘no’ were taken at face value. Responses provided the information to categorise participants into one of three groups. Participants who reported ‘no’ to both questions were categorised as non-fallers. Participants who reported ‘yes’ to a fall were considered fallers, regardless of any near fall occurrences. Participants who stated ‘yes’ to a near fall but ‘no’ to a fall were categorised as near fallers. Each participant was allocated to only one group. Details of each near fall or fall event were collected, including the cause and the total number of near falls and/or falls.

As previously stated, there were no established clinical tests to assess balance in midlife adults or to assess for near falls, so a new battery of tests was created. The starting point was a validated set of assessments for community dwelling older adults that incorporated both static and dynamic balance (Mackintosh et al., 2006). This core set of six tests consisted of standing feet together (FT), eyes open (EO) for 30 seconds; FT eyes closed (EC) for 30 seconds; single leg stance (SLS) right leg EO for five seconds; SLS left leg EO for five seconds; turn around to face 180°; and five tandem steps forward, starting and ending with feet together. As these tests had not been validated in adults younger than 65 years without known balance problems, further challenging tests were added. These incorporated functional activities.

The additional static tests had been validated across different age cohorts, including midlife and young-older adults (Springer et al., 2007). They comprised SLS EC for 5 seconds; SLS EO 30 seconds and SLS EC 30 seconds. The first additional dynamic test was tandem walk backwards, validated as a falls risk measure in young-older adults (Carter et al., 2019). The other functional dynamic tests, the FMS hurdle step and FMS lunge, were selected for their combination of dynamic movement incorporating strength and power, and their use in predicting balance outcomes in midlife adults (Harrison et al., 2021). For safety, the FMS lunge was originally performed on a mat on the floor and, if the participant passed safely, subsequently conducted on the FMS narrow beam as per protocol. Scoring for the FMS items were per protocol i.e., completed correctly = 3; completed with compensation = 2; unable to initiate or complete = 1; pain during test = 0 (Cook et al., 2006). The description of all tests and the pass/fail criteria for this study are provided in the manuscript (Baker et al., 2021a, Appendix C).

Associations between faller, non-faller and near faller groups were determined using independent samples Kruskal-Wallis tests. Chi-square tests for independence evaluated associations between group allocation, gender and test pass/fail results. Small (0.07), moderate (0.21) or large (0.50) effect sizes were determined (Pallant, 2020). The pass/fail status was compared between the three groups and results were explored with Odds Ratios, positive and negative predictive value, and area under the receiver operating characteristic (ROC) curve (AUC). Interpretation of odds ratio was taken as > 1 likely to fail the test, and <1 protective of failing test. The positive predictive value identified the probability of having experienced a near fall or fall. The negative predictive value identified the probability of not experiencing a near fall or fall. Wilcoxon Rank tests explored between group differences to produce the predictive capacity, sensitivity, and specificity. No adverse events occurred during balance testing (Baker et al., 2021a).

Of the 627 participants, the majority were non-fallers (n = 407, 64.9%). Near fallers constituted nearly one fifth (n = 121, 19.3%) and fallers the remainder of the cohort (n = 99, 15.8%). Fully detailed results are provided in the manuscript (Appendix C). Ages ranged from 40-75 years with similar age distribution across the non-faller, near faller and faller groups (non-fallers mean age 59.5y (+/- 10.6); near fallers mean age 60.7y (+/- 9.6); fallers mean age 61.6y (+/- 11.2)). There was no statistically significant difference between the groups for age (χ^2 (3, n = 627) = 4.18, p = 0.12). There were more female (77.0%) than male participants overall, with similar distribution across all three groups (χ^2 = 5.60, p = 0.06, Cramer's V = 0.06) (Baker et al., 2021a).

Overall, comparison of the balance test pass/fail results between near fallers and fallers showed no significant between-group differences, indicating that near fallers and fallers produced similar performances. In addition, near fallers and fallers were significantly more likely to fail the balance tests than non-fallers (Baker et al., 2021a).

Near fallers were significantly more likely to fail SLS EO 5 seconds (χ^2 = 12.3, p < 0.01, Cramer's V = 0.14), five tandem steps forward (χ^2 = 6.79, p = 0.03, Cramer's V = 0.10), the FMS hurdle step (χ^2 = 18.6, p < 0.01, Cramer's V = 0.17), the FMS lunge on floor (χ^2 = 16.0, p < 0.01, Cramer's V = 0.16) and the FMS lunge on beam (χ^2 = 13.8, p < 0.01, Cramer's V = 0.15) than non-fallers. The results are summarised in Table 3.2 Test results to distinguish near fallers from non-fallers.

Table 3.2 Test results to distinguish near fallers from non-fallers

Name of test	Odds Ratio (95%CI)	Significance (p)	Sensitivity	Specificity	Positive Predictive Value (%)	Negative Predictive Value (%)
SLS EO 5s	**2.7** (1.5-4.9)	0.00	18	92	40	79
Tandem steps forward x5	**2.5** (1.1-5.7)	0.02	8.9	96	40	78
FMS hurdle step	**2.9** (1.4-5.8)	0.00	13	95	43	78
Lunge on floor	**2.5** (1.5-4.1)	0.00	26	87	39	80
FMS lunge on beam	**2.1** (1.4-3.2)	0.00	52	65	31	83

Key: EO eyes open; FMS Functional Movement Screen; s seconds; SLS single leg stance; *significance ≤ 0.05; **significance ≤ 0.01

When these tests were grouped together for pass/fail, there was a sustained, moderate predictive capacity to discriminate near fallers from non-fallers (AUC 0.61 (0.56–0.65), p < 0.05, sensitivity 72.7%, specificity 49.4%) (Baker et al., 2021a, Appendix C). Therefore, these tests provided the first insights to the balance ability of a group of near fallers. The near fallers were two to three times more likely to fail single leg stance, tandem steps, FMS hurdle step and FMS lunge than non-fallers. These results identified a new battery of tests to discriminate near fallers from non-fallers (Baker et al., 2021a).

Chapter Summary

In this chapter, the overarching aim was to identify the best available materials and methods to investigate near falls in midlife and older adults living in the community. The exploration of different instrumentation types identified that IMUs were appropriate, affordable and accessible for community balance testing. Further investigation into the accuracy of the IMUs revealed they provide valid and reliable measures of static and dynamic balance in healthy adults. Further, they reveal variations in sway not seen from clinical assessment. These findings extended confidence for the use of inertial sensors in this project's data collection. Finally, clinical balance assessments were explored for midlife adults and a new battery of clinical tests to identify near fallers was created. The new group of balance tests - single leg stance, tandem steps, FMS lunge and FMS hurdle step - provided measures to discriminate near fallers from non-fallers. The next chapter describes the application of the materials and methods for testing postural sway in near fallers, midlife and older adults living in the community.

CHAPTER 4 METHODS AND PROTOCOL

The previous chapter identified the clinical balance measures and assured the reliability and validity of inertial sensors to measure balance. Those findings informed the design of activities and equipment for the study. This chapter explains the theoretical approach, design, population of interest, intervention, outcome measures and statistical analysis plan of measuring postural sway in near fallers. The following research questions guided the methods:

- What are the contributing factors to near falls in midlife and young-older adults?
- What are the differences in postural sway between near fallers and non-fallers in this population?
- What happens to postural sway in midlife and young-older adult near fallers during distraction?
- What happens to postural sway in midlife and young-older adult near fallers during fatigued balance testing?
- What is the predictive capacity of sway to identify midlife and young-older adult near fallers?

These research questions informed the study that investigated the degree of sway during balance tasks as a prediction for near falls. The study was based on the TRIPOD (Transparent Reporting of a multivariable prediction model for Individual Prognosis Or Diagnosis) Statement (Collins et al., 2015) and reported against a 22-item checklist (Appendix D). The TRIPOD statement is a quality checklist aimed at transparency, reducing bias and guiding objective assessment during the development, validation or update of a prediction model (Collins et al., 2015; Moons et al., 2015). The emphasis of the statement was on assessment for prediction of future near falls or falls. The emphasis on future provided an opportunity to identify modifiable contributing factors. These may be alterable by behaviour or lifestyle change approaches that address the associated risk. The principle of taking multiple potential contributing factors and assigning each a ‘weight’ to calculate a likelihood of the diagnosis occurring answered the first of the research questions.

The TRIPOD model uses a development framework, where relevant predictors are combined into multivariate models - logistic regression for short term and Cox regression model for long term. The following describes this study’s alignment with the TRIPOD short term model methods and plan for analysis.

Study Design

This longitudinal prospective cohort study consisted of three main components - a survey, balance testing and a near fall diary. The study was approved by the Human Research Ethics Committee at

Flinders University (#4084) (Appendix E). Four local councils and one aged care provider directly advertised the study, with further recruitment occurring through snowball sampling. The study flyer (Appendix F) was distributed in hard copy and online through local government and aged care provider centres, newsletters, social media, and service networks.

Population

The study recruited middle to young-older adults, aged 40-74 years, not previously investigated for, but potentially at risk of, future falls (Peeters et al., 2018; White et al., 2018). Exclusion criteria included current pain, trauma or severe pathology affecting ability to stand or walk; diagnosis of a neurological condition; cognitive or communication deficits that impeded ability to follow instructions safely; inability to walk 10 metres unassisted; joint surgery or injury within the previous 3 months; an unstable medical condition where active participation in exercise was contraindicated e.g. unstable cardiac condition, pulmonary embolus, stroke within three months (Fletcher et al., 2013); and allergy to sticking plaster (to hold the sensor in place) (see Appendix G Participant Information and Consent Form).

The recruitment materials included contact details to access the participant information and consent form. Participants who subsequently contacted the research team were screened by phone against selection criteria and a convenient testing time and venue were arranged. After providing informed consent, participants were directed to complete the electronic demographic and baseline questionnaire, described below. Participants were asked to wear loose comfortable clothing and appropriate shoes for exercising.

Setting

Eleven community venues were accessed for the balance testing component: four community centres at local government, four University venues, one retirement village, and the club rooms at a football club. Access to all venues was flat. Each venue had a room at least 12m long to incorporate the Incremental Shuttle Walk Test (ISWT) as described below.

Materials and Equipment

The following table (Table 4.1 Materials and Equipment) outlines the equipment required for the testing activities.

Table 4.1 Materials and equipment

Item	Purpose
Inertial Sensor	Measure sway
Digital scales	Calculate BMI
Stadiometer	Calculate BMI
Clinic clock with second hand	Time single leg stance
Pulse oximeter	Measure resting and exercise heart rates
Tape measure	Tibial length for lunge step
Tape for marking floor	Length of FMS lunge step; mark walking track
FMS rod	Stabilise arms during FMS lunge
iPad and laptop	Sensor data logging
Double-sided sticky tape	Hold sensor in place
Neoprene belt	Hold sensor in place
Audio for incremental shuttle walk test (ISWT)	Standard protocol for ISWT
Bluetooth speaker	Volume for audio track ISWT
Cones	Turnaround points ISWT
Printed rate of perceived exertion scale	Effort rate for ISWT
Printed recovery scale	Perceived recovery from ISWT
Randomisation sheet	Random order of tests and activities
Data collection sheets and pen	Write activity outcomes
Printed copies of diaries	To record near falls and falls for 3 months
Hand gel, disposable face masks, disinfectant	Hygiene before, during and after participants
COVID-safe plan	Plan provided to each council and venue

The balance tests were informed by the earlier study described in Chapter 3 and provided as Appendix C. The test protocol is outlined in Table 4.2 Balance tests.

Table 4.2 Balance tests

Balance test	Verbal Instruction	Pass/Fail criteria
Single Leg Stance 	*Stand on your preferred leg for 5 seconds with your eyes open.*	A ‘pass’ was considered standing unsupported on the leg of choice for 5 seconds, measured by second hand on the clinic clock. A ‘fail’ was considered four seconds duration or less, or if the standing leg moved its base of support, or if the participant touched an external support with hands or other body part.
Tandem steps 	*Start with your feet together, take five steps heel-touching-toe. Finish with your feet together.*	A ‘pass’ was considered five consecutive tandem steps. A ‘fail’ was considered less than five steps, any pivot of the feet, any sidestep or any external support.
Lunge 	*Hold the pole with both hands and position it along your spine against your head and down your back. Step forward on the floor with the preferred leg, placing the heel to the indicated mark. Lower the back knee to touch the floor gently behind the front heel. Return to starting position.*	FMS conventions 3 = performs the movement correctly without compensation; 2 = completes the movement but compensates in some way; 1 = unable to initiate or complete the movement pattern; 0 = participant experienced pain during the activity. Scoring in this study identified a ‘pass’ as FMS score 3 or 2. A ‘fail’ was an FMS score of 1 or 0.

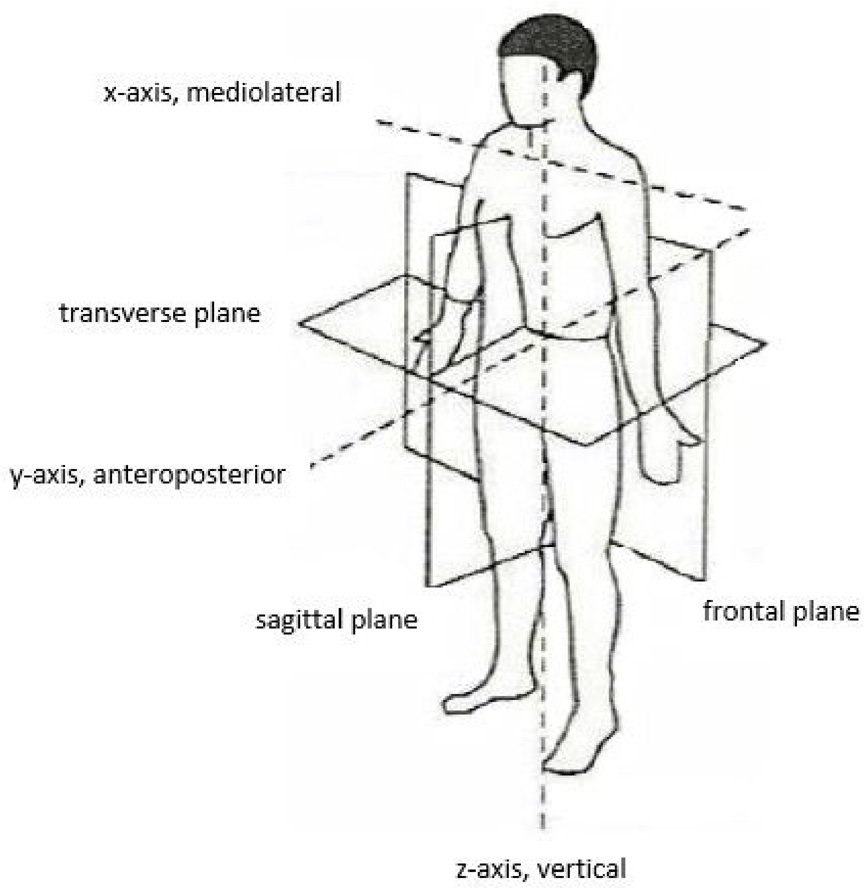
The step distance for the FMS lunge was calculated by the length of the participant's tibia (Cook et al., 2006). This study employed a tape measure instead of the usual FMS Hurdle Step measure because the tape measure is smaller and lighter for repeated transportation to different venues. The tibial length distance was noted then marked out on the floor with sticky tape. The first marker indicated the toe position of the standing foot. The second piece of tape marked the position for heel placement of the stepping foot, indicated by the tibial length. The other piece of equipment for the lunge was the rod, which was held behind the back, as per protocol (Cook et al., 2006) and outlined in Table 4.2 above.

A portable clinic clock with a second hand was purchased to time the single leg stance. Resting and exercise heart rates were measured using a portable pulse oximeter in the absence of chest strap heart rate monitors. Other important equipment included a stadiometer and digital scales for objective measurement of height and weight to calculate BMI. Overestimation of height and underestimation of weight in self-reported measures mean that objective measures of BMI are more reliable (Hodge et al., 2020). Personal protective equipment and cleaning equipment ensured that Covid-19 precautions were adhered to, in addition to usual clinical practice. A Covid-safe plan was provided to and approved by each site before recruitment took place (Appendix H Covid-safe plan).

The sensor chosen for this study was the Mbientlab MMR (Mbientlab, 2020) because it was affordable, lightweight, and rechargeable. This triaxial accelerometer and gyroscope combination has the same specifications as the more expensive versions such as APDM Opal identified in the systematic review (Appendix B, Table 3). The Mbientlab MMR was unobtrusive and lightweight, weighing 5 grams. It consisted of an accelerometer, gyroscope and magnetometer. The convenience of micro-USB recharging meant the sensors could be recharged overnight or after data collection periods. The sensor streamed data via Bluetooth to open-source App software for data acquisition installed on a laptop and an iPad.

The specific attributes of the sensor that were applicable for this research were gyroscope range +-125 – 2000 degrees/second, resolution 16-bit, sample rate 0.001Hz – 100Hz streaming, and accelerometer range +-2 - +-16g, resolution 16-bit, sample rate +/- 1300 μT (x, y-axis), +/-2500 μT (z -axis). The accelerometer provided data on directional changes due to movement. Three orthogonal axes provided accelerometry data. The x-axis provided acceleration data mediolaterally, the y-axis anteroposteriorly and the z-axis vertically (Figure 4.1 Sensor axes and planes aligned to the human body).

Figure 4.1 Sensor axes and planes aligned to the human body



(de Oliveira Sato et al., 2010. Reused under the terms of the Creative Commons Attribution Licence)

The three axes also aligned with the three orthogonal planes of the human body to orientate the embedded gyroscope. The x-axis aligned with the transverse plane which divided the body into upper and lower sections to describe flexion and extension movements. This plane provided pitch angular velocity forwards and backwards. The y-axis aligned with the sagittal plane, separating the left from the right side of the body. Data from the sagittal plane provided roll angular velocity, interpreted as side flexion or limb abduction movements. The z-axis aligned with the frontal plane, separating the front from the back of the body. This plane provided yaw angular velocity describing trunk rotation.

Additionally, the sensor contained a magnetometer with a 25Hz data rate. This was used as a marker for the start and end of each group of three balance tests by swiping a magnet past the sensor. Three MMR sensors were purchased. Each sensor was numbered and linked to either phone, laptop or tablet. Streaming was established with the accelerometer set to 200Hz, gyroscope set to 100Hz, and magnetometer set to 25Hz.

One sensor was used per participant and a new streaming session was initiated on the corresponding phone, tablet or laptop. Once the sensor and streaming device were synchronised, the sensor was placed on the participant over L4/5 (Patel et al., 2020) which was palpated through anatomical landmarks (see Figure 4.2 position of sensor on participant).

Figure 4.2 Position of sensor on participant

To prevent movement artefacts, the sensor was held in place by double-sided sticking plaster, with further support from a soft belt made of neoprene material.

Intervention

The first component of the study, the survey, gathered demographic and self-report information. Following this, balance tests were conducted while participants wore the inertial sensor. Balance was challenged under distracted and fatigued conditions. The balance tests provided a clinical pass/fail status and the sensor provided details of the sway during the activities. Finally, a daily diary captured any near fall or fall events for the three months after testing balance. While retrospective reporting of falls and near falls is easy to gather, it is prone to selective recall and potential bias from over- or under-reporting (Romli et al., 2021). Prospective falls and near falls reports sourced using a recognised method for data collection such as a diary, calendar or phone call can increase reliability of reports (Nagai et al., 2017). This study relied on retrospective reporting of no falls in the previous six months as the primary eligibility criterium. A prospective diary then provided the future near fall and fall events to determine group allocation for statistical analyses. The three components of the intervention are described in detail below.

1. Survey

The survey was provided electronically via the University Qualtrics online survey system, or in hard copy depending on participant preference (Appendix I). The unique ID was generated and entered by the participant. Demographic variables requested information on age and gender, whether the participant lived alone or with others (Petersen et al., 2020) and their paid or voluntary work status as employed, formal voluntary work or not in labour force (Australian Bureau of Statistics, 2021). The survey asked if the participant was afraid of falling (Johnson et al., 2019) and whether they had experienced any near falls in the previous three months. If the response to the latter was yes, further information on the number and cause of near fall was sought (Baker et al., 2021a).

To understand the self-perceived health of participants, validated surveys or their components were included. The SF36 quality of life survey sections on General Health (five questions), Vitality (four questions) and Physical Function (ten questions) incorporated self-reported quality of life and physical wellbeing, using the Rand scoring parameters (Hays et al., 1993).

For the sensory components, self-reports relating to hearing included whether the participant used a hearing aid (Riska et al., 2021), experienced tinnitus (Lastrucci et al., 2018), had a medical or surgical history of ear problems or ear infections (Heitz et al., 2019), or a history of noise exposure at work or socially (Mick et al., 2018). The speech and spatial qualities of hearing (SSQ5) was included to provide a subjective measure of hearing function (Demeester et al., 2012; Potts et al., 2019). Current experience of dizziness was collected and a 'yes' response directed the participant to complete the abbreviated dizziness questionnaire (Roland et al., 2015). Self-perceived quality of vision and distance vision was captured (Yip et al., 2014). A current history of wearing corrective vision aids (Ogliari et al., 2021b) was also collected, and if the response was yes, whether the glasses or contact lenses were bifocal or multifocal (Mehta et al., 2021). The final vision-related question was whether the participant had undertaken a vision test in the previous 12 months (Moore et al., 2011).

2. Balance Tests

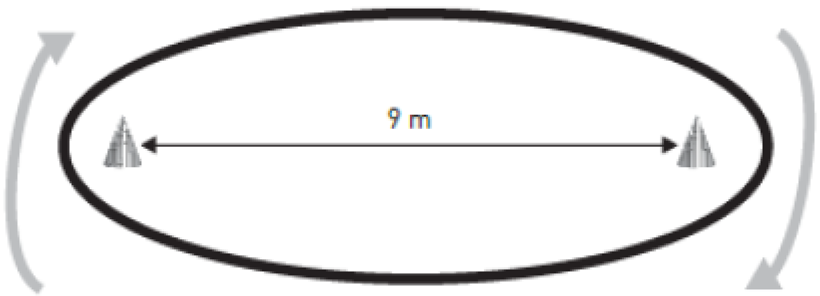
The balance tests that differentiated near fallers from non-fallers were chosen as activities to measure postural sway (Baker et al, 2021a, Appendix C). At the start of testing, the researcher described and demonstrated each of the balance activities. Participants were invited to practice each of the balance activities and to identify a preferred limb for standing on single leg and the stepping component in the lunge. The dominant limb was not considered to have a role in the balance outcome (Huurnink et al., 2014; Schorderet et al., 2021) and single leg stance with preferred choice of foot placement has been shown to improve stability (Gibbons et al., 2019). The lunge was taken from the FMS protocol (Cook et al., 2006) as described in Table 4.2 Balance tests, p.27. The participant's tibial length was attained by measuring from the top of the tibial tuberosity to the floor by the instep with shoes on. This distance was marked on the floor by two pieces of tape. As per FMS protocol, a rod was held behind the back touching the head, thoracic spine and sacrum. This precluded using arms for balance and raises the centre of mass. All participants first practised the lunge movement without the FMS rod but with their hands on their head, i.e., hands were able to be used for balance recovery if required. Using clinical judgement, participants who failed the preliminary lunge test without the rod were not offered the rod and achieved a maximum score of ‘1’. If the participant passed the lunge without the rod, they were then offered the rod for the complete movement.

Balance testing was further challenged by two confounding conditions: fatigue and distraction.

Fatigue

Physical fatigue in the lower limb muscles was induced through the Incremental Shuttle Walk Test (ISWT) (Probst et al., 2012). The ISWT is a maximal, externally paced field test conducted on a 10-metre walking track. It provides a standardised, validated test of physical fitness for community dwelling young, midlife and older adults (Agarwal et al., 2016; Dourado & Guerra, 2013; Harrison et al., 2013; Lima et al., 2019). The 10-metre walking track was set up according to protocol with cones set 9m apart to allow the last metre as the distance walked around the cone. Markers were set 0.5m inside each cone to identify the cut-off distance (see Figure 4.3).

Figure 4.3 Set-up for incremental shuttle walk test



Participants walked according to pre-recorded, timed beeps played audibly via Bluetooth speaker. The audio was the standard ISWT track purchased from University of Leicester Teaching Hospitals, UK, the owner of the ISWT. All testing was conducted inside buildings to provide consistent walking environments that were not dependent on the weather. Instructions for the ISWT were consistent and only the standardised phrases were used during the test (Appendix J). The test was terminated when the participant failed to reach the cone/marker in the time allowed, or when any of the following occurred per protocol:

- The participant was more than 0.5 metres away from the cone when the bleep sounded - one lap to catch up was offered
- The participant reported that they were too breathless to continue
- 85% of predicted maximum heart rate was attained
- The participant exhibited chest pain, acute confusion, loss of coordination, light-headedness, or leg cramps

As soon as the test ended, exercise heart rate was recorded via pulse oximeter, and the perceived effort was recorded via the modified Borg scale (exertion 6-20) (Appendix K) (Arney et al., 2019; Borg, 1982). Recovery was measured both objectively by heart rate recovery (oximeter) and subjectively by the Perceived Recovery Scale (Laurent et al., 2011, Appendix L). Heart rate readings were taken every minute for six minutes, or until the reading was within 10% of the resting heart rate. Concurrently, participants were asked to provide the corresponding perceived recovery score. An *a priori* decision was made to stop measuring heart rate recovery after 10 minutes if the perceived recovery score was ≥8/10 (Laurent et al., 2011).

Distraction

The second confounding condition was distraction. Two distraction tasks were provided in this study - categorical naming and serial subtraction. To provide equity for participants with strengths in either words or numbers, all participants were provided with both a categorical naming task and a subtraction task. Both categorical naming tasks and serial subtraction have been identified as appropriate distraction tasks to confound balance (Fallahtafti et al., 2020).

The categorical naming tasks were chosen because verbal fluency dual tasks identify significant differences between groups of fit and healthy young and middle-aged adults (Hadad et al., 2020). Categorical naming activities were generated for 25 categories, chosen from a free speech therapy website (Home Speech Home, 2021) and approved by a senior Speech Pathology clinical and teaching colleague. The categories comprised words beginning with the letters of the alphabet A, B, C, D, F, G, L, M, N, P, R, S, T, or W, and words relating to the topic of animals, body parts, clothes, colours, drinks, food, gardening, music, sports, toys, or transport. For the categorical task, the instruction to participants was consistent:

"Name as many **words** as you can…starting with the letter…

or …that relate to… (*animal, body part etc*)"

If participants commented on the difficulty of the categorical naming task when English was not their first language, leeway was provided through the research team supplying a different letter or category for the second and third balance tests (single leg stance or lunge or tandem) within that categorical naming activity.

The second distraction task was serial subtraction, identified to rely on working memory and executive function (Brustio et al., 2017). The use of serial subtractions for distraction tasks during balance has traditionally been undertaken during automatic balance functions such as walking and the Timed Up and Go Test (Fallahtafti et al., 2020). The rhythm of walking is meant to support serial subtraction, explained by common neural links between rhythm and verbal fluency (Fallahtafti et al., 2020). In this study, serial-7 subtraction was trialled. However, the non-automatic nature of the two dynamic balance tests chosen for this study - lunge and tandem walk - identified that serial-7 subtractions were too difficult. Practice attempts in the trial period before data collection demonstrated either standing still while subtracting by 7s or moving without subtracting by 7s. Therefore, serial-3 subtraction was introduced (Brustio et al., 2017). Further, when participants were unable to subtract by threes, they were instructed to subtract by twos, commencing on an odd number. If this was still not possible, they were directed to count backwards during the balance task. The focus of the task was to distract from balancing, not to measure language or arithmetic ability. The responses to the spoken distraction activity were not recorded.

Randomisation

Each participant was provided with a randomly generated, individualised order for tests and conditions, to prevent performance bias (Pannucci & Wilkins, 2010). Random orders were created for individualised categorical naming task and subtraction starting number as follows (Appendix M):

- balance tests (single leg stance (S), tandem (T) and lunge (L))
- first intervention (distraction or fatigue)
- categorical distraction task (letter or category)
- subtraction starting number

A list was created for permutations of balance testing order. The six options (S-L-T; S-T-L; T-L-S; T-S-L; L-T-S; L-S-T) were randomised using computer-program random list generation (Random Lists, n.d.). The 25 categorical naming options listed above were randomised on the same website. Similarly, the serial-3 subtraction, starting with a number between 100 and 399, were also randomised. Next, the first intervention (distraction or fatigue) was entered under a column named ‘condition order’. Following this, the first distraction task (starting number or category) was entered under the column ‘distraction alone’. Finally, the second distraction task i.e., the remaining number or category, was added under the column ‘distraction together’ to inform which aligned concurrently with the fatigue. This order was entered on the front page of the data collection sheet

(Appendix N) when the participant arrived for testing. Each participant completed the three balance tests - SLS, lunge and tandem - five times: baseline, distraction or fatigue, the alternate distraction or fatigue, concurrent fatigue with distraction, final testing. The whole session took approximately one hour for each participant.

As soon as the heart rate and Borg scale were recorded, the participants performed the three balance tests in their individually allotted order. Following the three balance tests, participants sat down and were provided with cold water refreshment.

3. Diary

At the end of the physical balance testing session participants were provided with a diary (Appendix O) either as paper copy or electronically, depending on participant preference. The diary captured the number and reason for any near falls or falls for the three months following the balance tests. Daily entries were encouraged. Reminders for diary completion were generated and sent once each week, usually by text message (Teister et al., 2018) or email if preferred. Reminders (Appendix P) included general information about balance and falls provided in succinct phrasing not intended to change participant behaviour (Kocielnik & Hsieh, 2017; Shimoni et al., 2020). Text messages reminders were sent by TextMagic online marketing software (TextMagic, 2001). At the end of each month, diaries were collected for data entry into the master data collection spreadsheet (Teister et al., 2018).

Data Capture

Each participant was asked to create a unique identity (ID) which was used for matching the survey, face-to-face testing, sensor data capture, and diary. If participants had not yet completed the survey when they attended for face-to-face testing, they completed a paper copy after the testing (Appendix I) before leaving the test venue. All survey data were entered in an Excel spreadsheet, kept on the University's secure research drive. Hard copies were kept in a locked drawer in the University.

At testing appointments, written data were collected on a purpose-built data collection sheet (Appendix N). The data sheet recorded the order of balance tests, the order of first intervention (distraction or fatigue) with the relative categorical naming task or serial-3 subtraction starting number, and the alternate categorical/subtraction task to be completed concurrently with the second fatigued testing. Resting heart rate was measured by pulse oximeter on the middle finger of the preferred hand before any balance or walking activity (Yuda et al., 2020). All balance activity data were captured on this form i.e., the commencement time of each set of three balance tests, the clinical pass or fail result for each test, the shuttle walk test level/stage completed on both occasions, the Borg exertion score, exercise heart rate, recovery heart rate at one-minute intervals and Laurent's perceived recovery scale at one-minute intervals. Any anomalies were noted during testing, such as a participant spontaneously attempting a test twice.

Data Extraction

The sensor data extraction was multi-stage and informed through consultation with biomedical and engineering colleagues and recent publications (Alsubaie et al., 2019; Lockhart et al., 2019; Soangra & Lockhart, 2018). The magnetometer .csv files were interrogated to identify peaks for start and end of each set of tests. These were cross matched with the start times for the balance tests written in the data recording sheet. Where this did not work, for example when the magnetometer did not record a swipe with the magnet, peak detection algorithms were written and applied in an attempt to identify the balance tests within the data. This method was not useful to differentiate the balance tests. Finally, the most consistent method was to visually inspect each participant's graph, cross reference with the order of tests (e.g., 1. Lunge, 2. SLS, 3. Tandem), the order of conditions (distraction or fatigue first) and the pass or fail for each test for each participant. Two people each independently extracted the first five full datasets containing all five conditions with three tests each (n = 75 tests). The start and end times for each of these tests were compared for accuracy and provided 73.3% agreement. Discrepancies were discussed in relation to the pass/fail status of the test, test anomalies such as the participant repeating a balance test unrequested, and the timing of tests, to differentiate the tests from other incidental activity such as walking to the start of the shuttle walk circuit or walking and turning to sit down to rest after the fatiguing activities. A second round of extraction comparing n = 30 tests provided 96.7% agreement.

Once the test orders and success of each test was known, each balance test was manually extracted from the accelerometer and gyroscope data and named as follows:

[sensor type][participant ID]_[condition]_[test]_[axis]_[measure]

Sensor type - a (accelerometer) or g (gyroscope)

Participant ID

Condition time - baseline (BL), distraction (DT), fatigue (shuttle walk - SW), both (Bth) or final (Fin)

Test (lunge (L), single leg stance (S) or tandem (T))

Axis (x, y, or z)

Measure e.g., min (minimum), max (maximum), ABM (absolute maximum) etc.

To interrogate each axis of each graph, the data were displayed as a scatter graph. For each test on each axis, the minimum and maximum acceleration (metres/second) or rotational velocity (degrees per second) was identified in MATLAB. These data were recorded in a spreadsheet for each test (Lunge, SLS and Tandem) under each condition (baseline, distraction, fatigue, fatigue with distraction, final). The absolute maximum values were identified to provide the peak acceleration or peak rotational velocity for each axis, condition and test (Ponciano et al., 2020).

At the completion of the testing session, the captured data were labelled with that participant's unique ID. Once labelled with the participant ID, the data were uploaded to the University cloud via Bluetooth, or the primary researcher's university email via Mail Drop to the cloud. In the cloud, the sensor data were available as comma separated value (.csv) spreadsheets. Each row of the spreadsheet corresponded to a time sample and the columns provided acceleration in metres per second (accelerometer) or rotation in degrees per second (gyroscope) for each of the three axes (x, y, z axes). Each complete .csv file was graphed in MATLAB – see sample graphs Figure 4.4 Accelerometer and 4.5 Gyroscope.

Figure 4.4 Sample full dataset - accelerometer

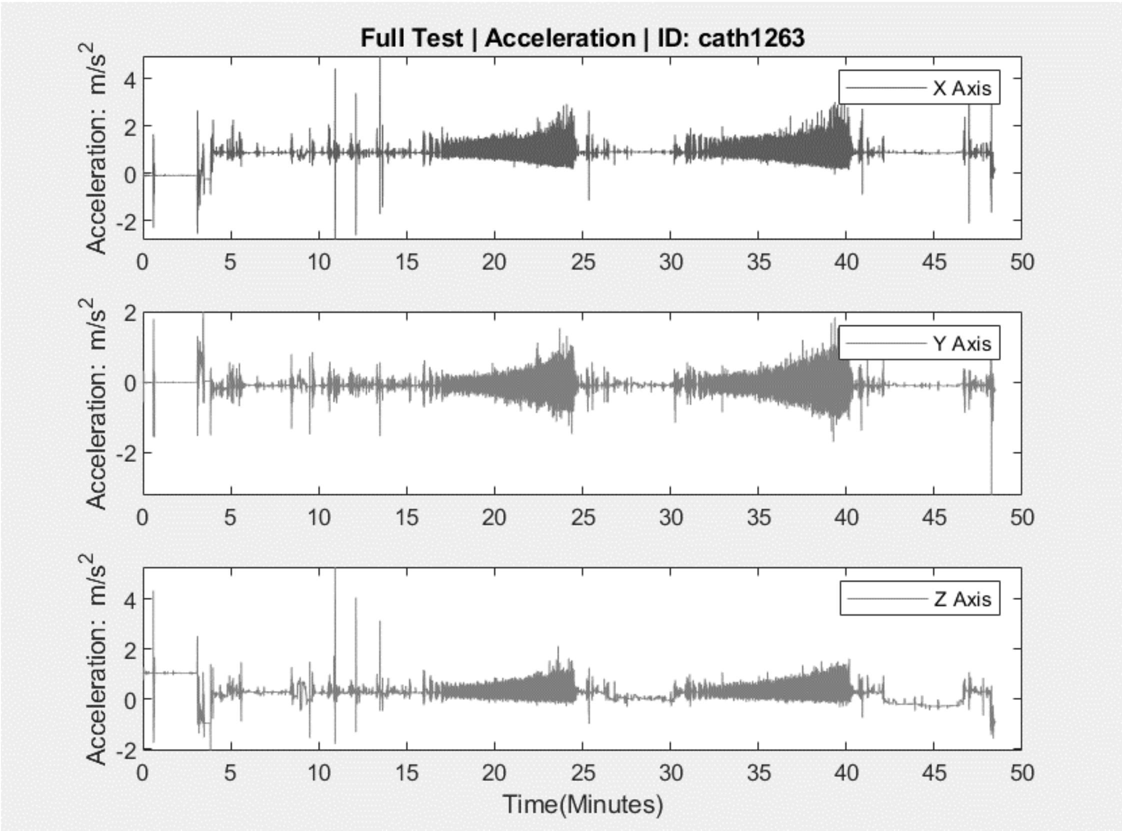
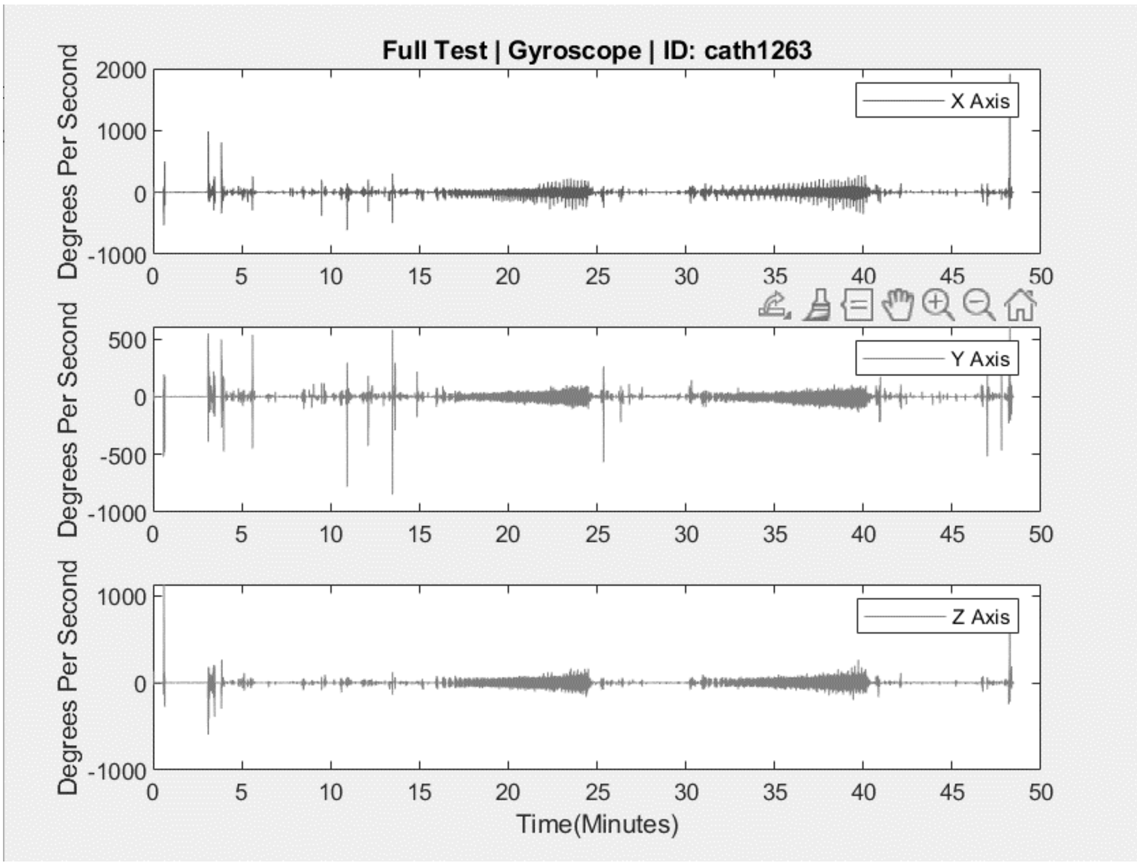


Figure 4.5 Sample full dataset - gyroscope



Sensor data analysis

The accelerometry data were analysed by root mean square (RMS) acceleration and vector magnitude acceleration. Acceleration vector magnitude and root mean square (RMS) acceleration in the mediolateral, anteroposterior and vertical directions have been identified as reliable measures of postural stability in healthy older adults (Hsieh et al., 2019). The RMS provided a gauge of the data distribution comparative to zero and was interpreted as the average magnitude in each direction (Soangra & Lockhart, 2018). The RMS for each axis on each test was calculated by squaring each result, summing the results, dividing by the number of cases, then taking the square root where n is the number of variables and i represents time:

$$RMS = \sqrt{\frac{1}{n}\sum_{i} x_i^2}$$

Acceleration vector magnitude (RMST) was interpreted as the magnitude of the whole movement, being a sum of squares of each axis (Soangra & Lockhart, 2018). Acceleration vector magnitude was calculated as

$$RMST = \sqrt{RMSx^2 + RMSy^2 + RMSz^2}$$

For the gyroscope data angular velocity about the pitch, roll and yaw planes was considered a valid and reliable measure of both static (Alsubaie et al., 2019) and dynamic (O'Brien et al., 2019) postural sway. In this study, the maximum absolute values of rotational velocity (degrees/second) were converted to angular velocity (radians/second). Radians (θ) were calculated as

$$\theta = degrees * (\frac{\pi}{180})$$

Angular velocity (ω) was calculated using the formula

$$\omega = \theta / t$$

Outcomes

The primary outcome of the study was the predictive capacity of sway to identify near fallers. Secondary outcomes were the relationships between the demographic variables and near falls, the differences between non-fallers and near fallers, and the changes in postural sway due to physical fatigue and distraction.

Statistical Analysis Plan

For retrospectively reported near falls via the survey, participants were allocated to one of two groups - retrospective near fallers (RNF) and retrospective non-fallers (R-non). Known fallers had been ineligible to participate. Near falls reported prospectively via the diary produced three groups: prospective faller (PF); prospective near faller (PNF); and prospective non-faller (P-non). As previously described, participants who sustained neither near fall nor fall were nominated non-fallers; those who experienced a near fall, but no fall were near fallers; and those who fell, regardless of near falls, were considered fallers. Both retrospective and prospective analyses investigated differences between near fallers and non-fallers only. Participants who reported prospective falls were excluded from further analyses.

Demographics

For the demographic data, categorical variables were dichotomised to provide binary responses. In these cases, a clinically relevant weighting was provided according to the relevant literature, as per TRIPOD statement (Collins et al., 2015). The more clinically relevant response aligned with poor postural sway or falls risk was provided with weighting of '1' and the more protective response was weighted '0' (Pallant, 2020). Weighting for categorical variables are summarised with the relevant reference in Table 4.3, Scoring Parameters.

Table 4.3 Scoring parameters for categorical demographic variables

Categorical Variables		
Variable	**Score weighting**	**Reference for weighting**
Near fall status	Near fall (1) No near fall (0)	Baker et al., 2021a
Gender	Female (1) Male (0)	Peeters et al., 2019
Age	Older adult (age 65-74) (1) Midlife adult (age 40-64) (0)	Duck et al., 2019
Living status	Living alone (1) Living with partner/others (0)	Petersen et al., 2020
Work status	Not in the workforce (1) Paid work, voluntary work (0)	Arpino & Solé-Auró, 2019
Vision quality	'Fair' or 'poor' (1) 'Good' or 'excellent' (0)	Yip et al., 2014
Vision test in past year	No (1) Yes (0)	Moore et al., 2011
Wear glasses or contact lenses	Yes (1) No (0)	Ogliari et al., 2021b
Bifocal/multifocal lenses	Yes (1) No (0)	Mehta et al., 2021
Distance vision	Fair/poor (1) Good/excellent (0)	Yip et al., 2014
Fear of falling	Yes (1) No (0)	Johnson et al., 2019
Hearing aid use	Yes (1) No (0)	Riska et al., 2021
Medical or surgical history affecting ears	Yes (1) No (0)	Heitz et al., 2019
History of ear infections	Yes (1) No (0)	Heitz et al., 2019
Noise exposure	Yes (1) No (0)	Mick et al., 2018
Tinnitus	Yes (1) No (0)	Lastrucci et al., 2018
Hearing disability (SSQ5)	Hearing disability ≤6 (1) No hearing disability ≥7 (0)	Potts et al., 2019
Dizziness	Yes (1) No (0)	Roland et al., 2015
BMI	Underweight or obese (1) Healthy or overweight (0)	Ogliari et al., 2021a
Balance tests	Fail (1) Pass (0)	Baker et al., 2021a

Key: BMI body mass index; SLS single leg stance; SSQ5 speech and spatial qualities of hearing

These variables were initially described as frequencies and percentages. Associations to near fall status were measured by Chi-square with Yates' Continuity Correction for 2x2 matrices and Phi or Cramer's V for effect size (Pallant, 2020). Effect size was interpreted as small (0.01-0.29), medium (0.30-0.49) or large (≥0.50) (Cohen, 1988). An *a priori* significance for all analyses was set at $p \leq 0.05$.

Continuous data consisted of age, height, weight, body mass index (BMI), resting heart rate, SF36 subscales for general health, vitality and physical function, and all the sway data. The SF36 subscales were scored according to the Rand instructions (Hays et al., 1993) where unanswered items were not imputed i.e., subscales were averaged according to the number of responses. Higher scores indicated better health status (Hays et al., 1993).

The BMI was calculated as weight in kilograms divided by the square of the height in metres and interpreted by Australian Government classifications of < 18.5 kg/m^2 underweight; 18.5 - 24.9 healthy weight; 25 - 29.9 overweight, ≥ 30 obese (Department of Health, 2021).

Baseline testing

All continuous variable distributions were assessed for normality by Kolmogorov-Smirnov statistics (Pallant, 2020). As no continuous variable was normally distributed, non-parametric statistics were used throughout. The results were reported as the standardized test statistic (*z* score), significance (p) and effect size (Pallant, 2020). The entire cohort balance test differences (lunge, SLS and tandem) were investigated with Friedman's test, with Bonferroni adjusted significant values for multiple tests (Field, 2018). Likelihood of future falls in the retrospective near fall group were assessed with odds ratios, positive and negative predictive values, and area under the curve.

Distracted and fatigued testing

The changes in sway from baseline to distracted and fatigued testing across the entire cohort were analysed using Wilcoxon Signed Rank tests. The differences between near fallers and non-fallers for all tests and under all conditions were analysed using Mann Whitney U tests.

Regression analyses

Finally, a binary logistic regression model evaluated the association between the variables and prospectively reported near fall outcome . Prediction development was through a Wald forward stepwise binary logistic regression model. Variables were grouped as follows, using the outcome, risk rating and reference outlined in Table 4.3 above:

- Demographics – self-reported descriptors of the participant cohort
    - Age, gender, living status, work status

- Biometrics – measures of participants' physical characteristics
    - Resting heart rate, body mass index, hearing, vision, dizziness
- Experiential - self reports of emotion and self-perceived function
    - Fear of falling, SF36 general health, SF36 physical function, SF36 vitality
- Sway data for the lunge, single leg stance and tandem steps
    - Vector magnitude acceleration, RMS acceleration (mediolateral, anteroposterior, and vertical), angular velocity (pitch, roll and yaw)

Near fall status was used as the dependent variable. Spearman correlation coefficients were generated to assess the strength of the relationship between the continuous predictor variables. A strong correlation coefficient of ≥0.8 indicated multicollinearity (Field, 2018, p. 402) and these variables were excluded. Next, univariate analyses were ranked on the strength of their association with the outcome of near falls, using the odds ratios and narrow confidence intervals that did not include 1.0 (Field, 2018). The stepwise logistic regression model used the strongest predictors. These were added to the model in turn, and the changes in -2 Log Likelihood, corresponding chi-square test and significance level were noted each time for changes in variance. Predictions for future near fall were further assessed using receiver operating characteristic (ROC) curves, the area under the curve (AUC) with 95% confidence intervals, sensitivity and specificity. The positive predictive value was calculated to provide the percentage correctly classified as near faller, and the negative predictive value to correctly classify the non-fallers.

The following research questions (RQ) indicated the order for analyses:

RQ1 what contributing factors are associated with near falls? Hypothesis (HO)1 Near falls are associated with older age, fear of falling, hearing impairment, BMI indicating underweight or obese, dizziness, reduced vision, poor general health, low vitality and poor physical function.

RQ2 Are retrospective near fallers more likely to fall prospectively than the retrospective non-fallers? HO2 participants reporting near falls retrospectively will report a fall by the end of the data collection period.

RQ3 What were the baseline differences in sway between near fallers and non-fallers? HO3 Near Fallers will demonstrate increased sway compared to non-fallers.

RQ4 What is the change in sway from baseline to distracted conditions? Is there a difference between near fallers and non-fallers under distracted conditions? HO4 Sway increases under distraction compared to baseline. Near Fallers will have increased sway compared to non-fallers during distracted testing.

RQ5 What is the change in sway from baseline to fatigued conditions? Is there a difference in response to fatigue for near fallers compared to non-fallers? HO5 Sway increases under fatigued conditions compared to baseline. Near fallers will have increased sway compared to non-fallers during fatigued tests.

RQ6 What is the predictive capacity of sway to identify near fallers? HO6 Acceleration vector magnitude, RMS acceleration and/or angular velocity outcomes will predict who will be a near faller.

Sample

Sample size was calculated for a prospective cohort design, using ROC curves and AUC, informed by the clinical balance test data (single leg stance, tandem walk, lunge) between near fallers and non-fallers (Baker et al., 2021a). In that study, the battery of clinical tests to discriminate the near fallers from non-fallers resulted in an AUC of 0.61, i.e., moderate discrimination. This study was designed to measure sway using accelerometry data from the inertial sensor. In a study using inertial sensors to identify fallers from non-fallers in a group of older adults undertaking a dual task functional walking test, the AUC for Timed Up and Go was 0.683 - 0.840 (Ponti et al., 2017). In a similar study for static postural sway, AUC of 0.745 - 0.755 was established (Hsieh et al., 2019). Therefore, using the conservative AUC from inertial sensors of 0.683 for the power calculation in this study provided a sample size of 112 participants (26 positive and 86 negative cases). In addition, due to potential dropout from prospective diary entries of up to 29.1% (Tan et al., 2018), the sample size of 112 was considered only 70.9% of those required. Therefore, the study sample size was calculated as 158 participants.

For the regression analyses, the sample size informing reliable outcomes was taken from the formula ($n > 50 + 8m$) where m corresponded to the number of predictor variables entered into the model (Tabachnick and Fidell, 2013, p. 123). Therefore, for a sample size of 158, the maximum number of predictor variables able to be included in the model was 13.

Data Management

The data collected from survey, sensor and face-to-face testing data were matched using the participant's unique ID. Data were entered into a spreadsheet on the Flinders University research drive, kept secure by password protection and encryption. All written data were transferred to the electronic master data collection spreadsheet and cross checked for accuracy. Paper copies of surveys and balance testing results were locked in a drawer in the University building. Signed consent forms were kept locked in a separate area in the university to prevent any possibility of re-identification.

Analysis of Reasons for Near Falls

The reasons for near falls pre-testing were entered as free text in the survey and in diary entries after testing. Each near fall incident was entered into a spreadsheet with the description of cause written verbatim from the survey and diary responses. The number of incidents was also recorded, to ensure each near fall event was described. Three schemes were trialled on the retrospective survey reports to explore the most consistent method for coding:

1. definition of near falls as *“slips, trips, stumbles, missteps, incorrect weight transfer or temporary loss of balance”* from Pang et al., (2019).
2. cause of near fall or falls categorised as intrinsic (cardiovascular, musculoskeletal, cognitive load, or other sensory cause) or extrinsic (uneven surface, wet surface, low light, external perturbation or clothing/shoes) described by Baker et al., (2021a).
3. according to the physical, behavioural or environmental risk factors for falls as described by the National Council on Aging (2016).

The process for coding the narrative involved three independent coders to reduce the reliance on personal interpretation (Syed & Nelson, 2015). Coding books were provided to the coders, Table 4.4.

Table 4.4 Coding Table

Definition of near fall	Code	Example from surveys or diaries
Pang et al., 2019		
Slips	PS	*Slipped on tiny gumnuts on path*
Trips/stumbles	PT	*Caught my toe while walking*
Missteps	PM	*Looking up*
Incorrect weight transfer	PI	*Needed to take a longer step to balance*
Temporary loss of balance	PL	*Moving too fast*
Intrinsic (I-) or Extrinsic (E-)		
Cardiovascular system	ICV	*Getting out of bed too quickly*
Musculoskeletal system	IMS	*Ankle instability*
Cognitive load	ICL	*Inattentiveness*
Sensory system	ISS	*Suspected Meniere's*
Uneven surface	EUS	*Irregularities on footpath*
Wet surface	EWS	*Slipped on rocks at beach*
Low light	ELL	*Tripped over step in dark*
External perturbation	EEP	*Dog knocked me over*
Clothes, shoes	ECS	*Caught foot in long PJs*
National Council on the Aging, 2016		
Physical	PHY	*Vertigo*
Behavioural	BEH	*Didn't lift my feet high enough*
Environmental	ENV	*Rough surface when walking*

Each person coded the entire dataset of retrospective near fall incidents (n = 112) against the three methods. Percentage agreement was confirmed by intraclass correlation coefficient (ICC) which was excellent (ICC = 0.889, p = 0.00) for the first option (Pang et al., 2019). This result was more consistent than the intrinsic/extrinsic (ICC = 0.71) and physical/behavioural/ environmental (ICC = 0.61) methods for coding, therefore the first coding schema was used for both retrospective and prospective reports of near falls.

A ‘trip’ or ‘stumble’ of incorporated any description of the foot striking or catching against something. The reasons for the foot catching an object were considered any combination of: a) vision-related i.e., not seeing the obstacle due to low light, clutter, carrying items to block the view etc.; b) distraction or lack of attention; c) fatigue i.e., unable to lift the foot high or fast enough to clear the obstacle, including ‘tripping over own feet’; d) unanticipated movement of the tripping object e.g., pet moving into the path of movement; or e) intoxication. ‘Missteps’ were categorised by the prefix ‘mis-’ e.g., ‘misjudge depth of surface’, ‘missed the bottom step’ or ‘missing a rung when stepping off a step ladder’. ‘Temporary balance loss’ described external perturbations where the participant was hit by something unexpected e.g., ‘bumped by shopper in store’, as well as intrinsic causes such as ‘getting out of bed too quickly’ or ‘vertigo’. Weight-transfer errors included any phrasing that included ‘lost balance’, as well as descriptions of changing weight-bearing during a step e.g., ‘needed to take a longer step to balance’. ‘Slips’ were coded from occasions when the foot lost traction with the ground.

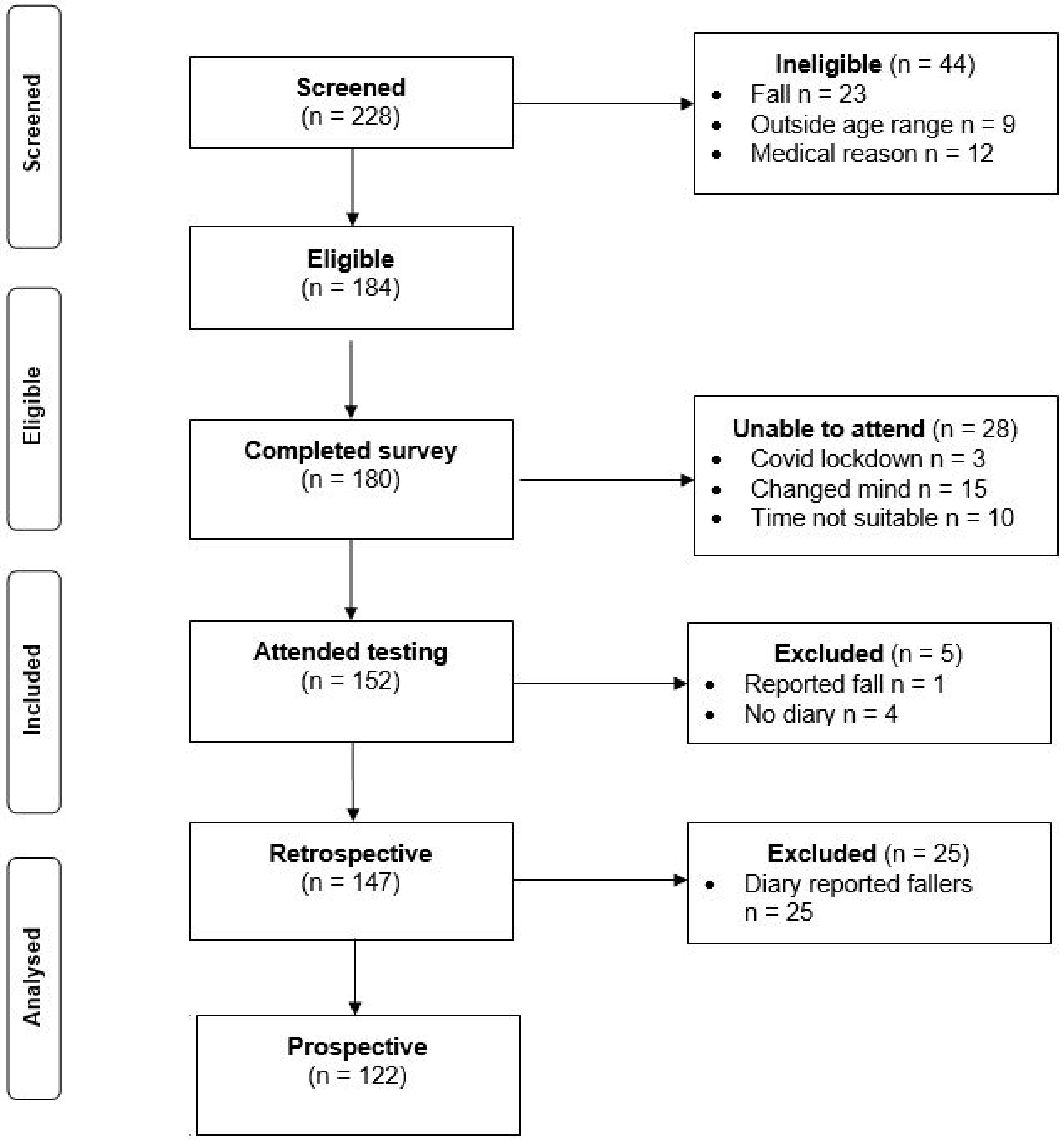
Chapter Summary

This chapter described the design, population, intervention, outcome measures and statistical analysis plan for the study. The next chapter describes the demographic results for near falls and explores the differences between retrospective and prospective reports of near fallers.

CHAPTER 5 DEMOGRAPHIC RESULTS

This study investigated postural sway in near fallers and non-fallers. Of an initial 223 community dwellers who indicated interest, 152 participants attended for face-to-face testing, and 147 were included in the demographic analysis - see Figure 5.1 Flowchart of Participants.

Figure 5.1 Flowchart of participants



Analyses were conducted to investigate relationships between self-reported near faller status and self-reported sociodemographic and objectively measured physical test results. Results are first described for the retrospectively reported near falls, where groups were allocated by survey responses. After this the prospectively reported near fall data, where groups were allocated by diary responses, will be explored. The comparison will provide useful information on the differences in results provided by the same group of people, to inform future data collection methods.

Retrospective Near Falls

Retrospective survey data provided two groups for comparative analysis: retrospective non-fallers (n = 109, 74.1%) and retrospective near fallers (n = 38, 25.9%). Participant characteristics are described in Table 5.1. Comparative analysis of retrospective near faller and non-faller characteristics.

Table 5.1 Comparative analysis of retrospective near faller and non-faller characteristics

Demographic	**Total n = 147** **median (IQR)**	**Retrospective near fallers** **n = 38 median (IQR)**	**Retrospective non-fallers** **n = 109 median (IQR)**	***Z* score (significance)** ***effect size***
Age *years*	64 (56, 69)	64 (54.8, 68)	64 (56, 70)	-0.44 (0.66) *0.00*
BMI *kg/m²*	26.1 (23.4, 29.3)	25.1 (22.6, 28.8)	26.6 (23.8, 29.3)	-1.23 (0.22) *0.01*
SF36 Gen Hlth	75 (60, 85)	70 (48.8, 80)	80 (65.0, 90.0)	**-2.89 (0.00) *0.02***
SF36 Vitality	70 (52.5, 80)	60 (45, 71.2)	75 (60.0, 80.0)	**-3.61 (0.00) *0.02***
SF36 Phys Fn	88.9 (77.8, 94.4)	86.1 (77.8, 94.4)	94 (77.8, 94.4)	-1.09 (0.27) *0.01*
Walk distance *m*	530 (415, 640)	543 (376, 646)	530 (433, 638)	-0.16 (0.88) *0.00*
Demographic	**Total** **n (%)**	**Retrospective near fallers** **n (%)**	**Retrospective non-fallers** **n (%)**	**Chi-square (significance)** ***effect size (φ)***
Gender *n female*	111 (74.1)	32 (84.2)	79 (72.5)	2.10 (0.15), *0.12*
Living alone	38 (25.9)	8 (21.1)	30 (27.5)	0.62 (0.43), *-0.07*
In workforce	64 (43.5)	17 (44.7)	47 (43.1)	0.03 (0.86), *0.01*
Fear of falling	48 (32.7)	15 (39.5)	33 (30.3)	1.08 (0.30), *0.09*
Hearing aid	12 (8.2)	5 (13.2)	7 (6.4)	1.71 (0.19), *0.11*
Med/Surg Ear	6 (4.1)	1* (2.6)	5 (4.6)	0.28 (0.60), *-0.04*
Ear Infections	19 (12.9)	6 (15.8)	13 (11.9)	0.37 (0.54), *0.05*
Noise exposure	13 (8.8)	3 (7.9)	10 (9.2)	0.06 (0.81) *-0.02*

Tinnitus	39 (26.4)	12 (31.6)	27 (24.8)	0.67 (0.41), *0.07*
Hearing disability	21 (14.3)	7 (18.4)	14 (12.8)	0.72 (0.40), *0.07*
Dizziness	46 (31.3)	18 (47.4)	28 (25.7)	**6.16 (0.01), *0.21***
Poor Vision	37 (25.2)	15 (39.5)	22 (20.2)	**5.57 (0.02), *0.20***
Distance Vision	28 (19.0)	11 (28.9)	17 (15.6)	3.26 (0.07), *0.15*
Vision test	27 (18.4)	6 (15.8)	21 (19.3)	0.23 (0.63), *-0.04*
Wear glasses	114 (77.6)	31 (81.6)	83 (76.1)	0.48 (0.49), *0.06*
Wear bifocals	71 (48.3)	23 (60.5)	48 (44.0)	3.07 (0.08), *0.14*
Prospective near fall	63 (42.9)	18 (47.4)	45 (42.2)	5.29 (0.07), *0.07*
Prospective fall	25 (17.0)	10 (26.3)	15 (13.8)	3.15 (0.07) *0.15*

Key: BMI body mass index; Gen Hlth general health; IQR interquartile range; Med/Surg ear medical or surgical history of problems with the ear; Phys Fn physical function; * cell size less than 5; Significant findings written in **bold.**

There were no significant differences for age, gender, living status, work status, fear of falling, BMI, reported hearing issues or tinnitus between the retrospectively reported near fallers and non-fallers. However, the retrospectively reported near fallers were significantly more likely to have self-reported poor vision than the non-fallers ($X^2 = 5.47$, $p = 0.02$, φ 0.20). This association did not extend to wearing multi- or bifocal glasses, or to poor distance vision. Near fallers were also more likely to report dizziness than non-fallers ($X^2 = 6.06$, $p = 0.01$, φ 0.21) but this did not extend to any of the refined questions regarding pattern, intensity, frequency or duration of dizziness. This suggests that people who recall a stumble or momentarily loss of balance are more likely to report poor vision or a problem with dizziness.

Retrospective near fallers had significantly poorer self-reported general health (z-score -2.89, $p = 0.004$, $r = -0.24$) than non-fallers. Similarly, retrospective near fall self-reports of vitality were significantly lower than the non-fallers (*z*-score = -3.61, $p < 0.00$, $r = -0.30$). These two results suggest lower energy levels and higher fatigue levels in the retrospective near fallers than their non-faller counterparts. Interestingly, there was no statistically significant difference in the physical function levels between the two groups, indicating that the physical function levels were similar between the retrospectively reported near fallers and non-fallers.

Investigation of the pass/fail status for the balance tests identified that there were no significant differences in pass or fail between near fallers and non-fallers in the lunge tests at baseline or under any of the challenging conditions of distraction or fatigue (Table 5.2 Balance test pass or fail status in retrospective near fallers and non-fallers).

Table 5.2 Balance test pass or fail status in retrospective near fallers and non-fallers

Test	Retrospective near faller (pass/fail)	Retrospective non-faller (pass/fail)	Chi-square	Significance p =	Effect size
Base Lun	26/12	77/32	0.07	0.80	0.02
Base SLS	33/5	97/12	0.13	0.72	0.03
Base Tan	27/11	83/26	0.39	0.53	0.05
DT Lun	27/11	79/30	0.03	0.87	0.01
DT SLS	30/8	91/18	0.40	0.53	0.05
DT Tan	27/11	79/30	0.03	0.87	0.01
SW Lun	26/12	82/27	0.67	0.41	0.07
SW SLS	28/10	95/14	**3.7**	**0.05**	0.16
SW Tan	23/15	78/31	1.60	0.21	0.10
Both Lun	26/12	78/31	0.13	0.71	0.03
Both SLS	27/11	95/14	**5.18**	**0.02**	0.19
Both Tan	26/12	73/36	0.03	0.87	-0.01
Final Lun	28/10	78/31	0.06	0.80	-0.02
Final SLS	33/5	98/11	0.27	0.60	0.04
Final Tan	29/9	84/25	0.01	0.93	0.01

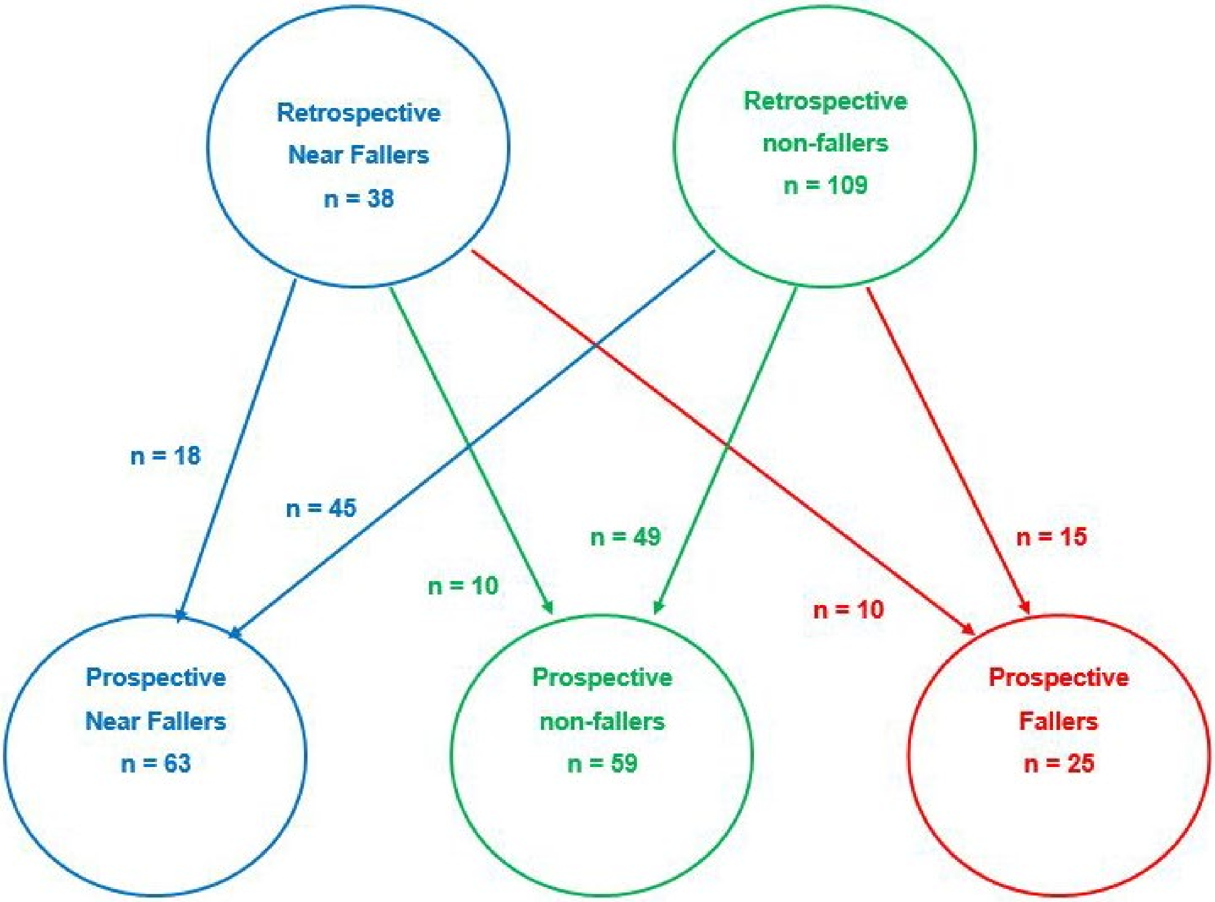
Key: Base baseline; Both distraction with fatigue; DT distraction; Lun lunge; SLS single leg stance; SW (Shuttle Walk) fatigue; Tan tandem steps

Retrospectively reported near fallers were significantly more likely to fail single leg stance under fatigued conditions (X^2 = 5.33, p = 0.02, Phi = 0.20). They were also significantly more likely to fail the single leg stance in the fatigued with distraction test ('both') (X^2 = 5.33, p = 0.02, Phi = 0.20) but not during distraction alone. These results indicated that retrospective near fallers' balance was significantly more affected by the fatiguing activity, which aligns with the SF36 results of significantly less vitality and general health in the near fallers. There were no significant differences between retrospectively reported near fallers and non-fallers for pass and fail results in the lunge or tandem steps at baseline or under any of the conditions.

The floor surfaces across venues consisted of wooden floors (n = 2), linoleum over cement floor (n = 4) and carpet over cement floor (n = 5). These surfaces did not influence the balance test results, as indicated by no significant differences in the chi-square analyses, ranging between 0.55 to 4.86 (Appendix Q).

Retrospective near fallers were as likely as the retrospective non-faller group to continue to experience near falls or falls in the following three months. The flow of participants from the retrospectively reported groups to the prospectively reported groups is outlined in figure 5.2 Retrospective to prospective group transitions.

Figure 5.2 Retrospective to prospective group transitions



Of the participants who retrospectively reported a near fall, almost half continued to prospectively report near falls (n = 18, 47.4%). They remained a near faller prospectively. Just over one quarter of retrospectively reported near fallers reported no prospective incidents, and therefore changed status to the prospective non-faller group (n = 10, 26.3%). The remaining quarter of retrospectively reported near fallers went on to fall in the next three months (n = 10, 26.3%) which changed their status to prospective fallers. While there was no significant association between retrospective near falls and prospective falls ($X^2 = 3.15$, $p = 0.07$), odds ratio calculations indicated that retrospective near fallers were 2.24 times more likely to have a fall than their non-faller counterparts (OR 2.24, 95% C.I. 0.91 - 5.53, $p = 0.08$). Although this result was not statistically significant and not a true representation of the population (wide confidence interval that crossed 1.0), the results were clinically relevant.

In the retrospectively reported non-fallers, almost half (n = 49, 45.0%) remained non-fallers based on prospective reporting, and retained their (prospective) non-faller status. However, almost the same number of retrospective non-fallers experienced a near fall incident in the following three months (n = 45, 42.2%). A small number of the retrospective non-fallers became prospective fallers (n = 15, 13.8%). None of these group relabelling produced statistically significant results to indicate a strong likelihood of near fallers becoming fallers, staying near fallers or having no further events.

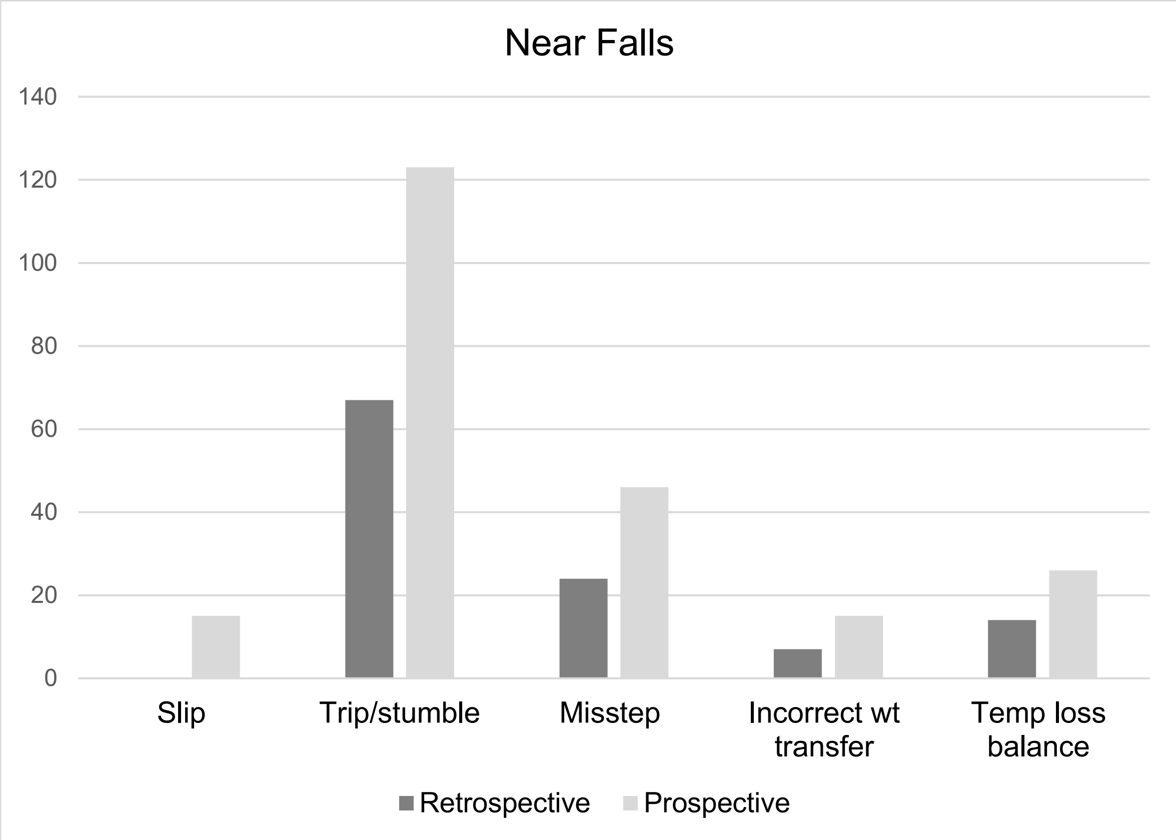
As previously described, the reliance on retrospective reports of falls can be inconsistent. Given the lack of event, embarrassment, or injury with a near fall instead of a fall, the reliance on retrospective reports of near falls is even more dubious. Therefore, to gather reliable information on the number and cause of near falls, the diary entries provided more trustworthy data. The next section investigates the demographic variables with near faller and non-faller groups allocated according to near falls reported prospectively, via diary entries.

For prospectively reported near falls, each participant was allocated to one of three groups according to their diary entries: prospective non-faller (n = 59, 40.1%), prospective near faller (n = 63, 42.9%), and prospective faller (n = 25, 17.0%). As the purpose of this thesis was to examine the differences between near fallers and non-fallers, the fallers were excluded from any further analyses. This resulted in a cohort of only near fallers and non-fallers, sample size of n = 122 (non-fallers n = 59, 48.4%; near fallers n = 63, 51.6%). All following statistical analyses were calculated on the cohort of prospectively reported near and non-fallers.

Reasons for Near Falls

The number of participants who reported a near fall increased with prospective reporting - from a quarter of the cohort retrospectively (25.9%) to just less than half the cohort prospectively (42.9%). Similarly, the number of near falls reported prospectively (n = 225) was twice the number reported retrospectively (n = 112). However, the mean number of near falls per person reduced from 3.1 (SD 1.6, range 1-6) retrospectively to 2.8 (SD 2.4, range 1-13) prospectively. Therefore, with the routine reminders and regular completion of the diary for prospective reporting, many more participants reported the occasional near fall, and fewer participants reported multiple near falls. The coding schema described in detail previously (Chapter 4 Methods p. 48) was applied to both the retrospective and prospective reports of near falls. Each near fall event was coded to either a slip, trip/stumble, misstep, incorrect weight transfer or temporary loss of balance. The results are depicted below (Figure 5.3 Near Fall Reasons).

Figure 5.3 Near fall reasons



Key: temp temporary; wt weight

While there were no reports of slips retrospectively, these incidents were recorded 15 times prospectively, but constituted less than 5% of all near fall events. Slip incidents were mainly due to wet or muddy surfaces, but slips were also coded for any lack of traction, e.g., caused by 'tiny gumnuts on the path' or 'loose gravel on slope'. Prospectively reported incidence of missteps (n = 46), incorrect weight transfers (n = 15) and temporary balance loss (n = 26) were twice as frequent as retrospectively reported incidents, aligning with the overall reporting pattern. Stumbles and trips, where the foot caught on an object, also nearly doubled (n = 67 retrospectively; n = 123 prospectively). These latter descriptors were the most common cause of all near falls, constituting over half (56.4%) of all events. Because they were the most common cause for near falls, the data were interrogated for patterns relating to the reason for the trip as previously described: not able to see the hazard; not noticing the hazard due to attention elsewhere; fatigue-related; or unexpected movement of the object. However, the limited detail provided in the 'reason' section of the diaries was not enough to be certain. For example, the reason provided, 'tripped uneven path', may have occurred because the participant did not see the raised section e.g., due to shadow, low light, poor vision etc., or not noticing the raised section i.e., due to focus elsewhere. Kerbs, paths and pavements caused almost one third of all trip events (32%). Certainly, the majority of the trip events occurred outside, including in gardens, on bushwalks and in the countryside. In these cases, the cause may conceivably have been fatigue, but this cannot be confirmed. Pets, particularly dogs and cats, created tripping hazards in 15% of the reports.

Prospective Near Falls

Associations between the demographic variables and prospective near fallers were investigated. For hearing-related medical/surgical interventions and noise exposure, numbers in the grid were too low to compute accurately and therefore were not reported (Pallant, 2020). The raw data are presented in Table 5.3. Comparative analysis of prospective near faller and non-faller characteristics.

Table 5.3 Comparative analysis of prospective near faller and non-faller characteristics

Demographic	**All participants**	**Near Fallers** **n = 65 (%)**	**non-fallers n = 59 (%)**	**Chi-square (significance),** ***effect size (Phi)***
Gender *n female*	93 (76.2)	49 (77.8)	44 (74.6)	0.17 (0.68) *0.04*
Living alone	34 (27.9)	21 (33.3)	13 (22)	1.94 (0.16) *0.13*
Not in workforce	54 (44.3)	27 (42.9)	27 (45.8)	0.10 (0.75) *-0.03*
Fear of falling	40 (32.8)	23 (36.5)	17 (28.8)	0.82 (0.37) *0.08*
Hearing aid	10 (8.2)	5 (7.9)	5 (8.5)	0.01 (0.91) *-0.01*
Ear Infections	14 (11.5)	9 (14.3)	5 (8.5)	1.01 (0.31) *0.09*
Tinnitus	35 (28.7)	16 (25.4)	19 (32.2)	0.69 (0.41) *-0.08*
Hearing disability	20 (16.4)	10 (15.9)	10 (16.9)	0.03 (0.87) *-0.02*
Dizziness	35 (28.7)	22 (34.9)	13 (22.0)	2.47 (0.12) *0.14*
Poor Vision	33 (27.0)	19 (30.2)	14 (23.7)	0.64 (0.42) *0.07*
Poor distance Vision	23 (18.9)	12 (19.0)	11 (18.6)	0.00 (0.96) *0.00*
No vision test	24 (19.7)	12 (19.0)	12 (20.3)	0.03 (0.86) *-0.02*
Wear glasses	93 (76.2)	49 (77.8)	44 (74.6)	0.17 (0.68) *0.04*
Wear bifocals	59 (48.4)	31 (49.2)	28 (47.5)	0.04 (0.85) *0.02*

Demographic	Total n = 122 median (IQR)	Near Fallers median (IQR)	non-fallers median (IQR)	*Z* score (significance), *effect size (r)*
Age *years*	64 (56,69)	64 (56, 69)	64 (55, 71)	-0.79 (0.43) *0.07*
BMI *kg/m²*	26.2 (23.5, 29.3)	26.1 (23.2, 30.4)	26.4 (23.5, 28.5)	0.18 (0.86) *-0.02*
SF36 General Health	75 (61.9, 87.5)	75 (60, 87.5)	75 (65, 90)	-0.28 (0.78) *-0.03*
SF36 Vitality	70, (61.9, 87.5)	70 (52.5, 80)	67.5 (57.5, 80)	-0.44 (0.66) *-0.04*
SF36 Physical Function	91.6 (77.8, 94.4)	94.4 (83.3, 94.4)	88.9 (66.7, 94.4)	1.05 (0.29) *0.09*
Walk distance *metres*	510 (400, 630)	520 (425, 640)	510 (405, 635)	0.45 (0.65) *-0.04*

Key: BMI body mass index; IQR interquartile range

The demographic variables and the balance test results showed no significant differences between the prospectively reported near fallers and non-fallers. These results rejected the first hypothesis, that near falls were associated with older age, fear of falling, hearing impairment, BMI indicating underweight or obese, dizziness, reduced vision, poor general health, low vitality, and poor physical function. The results also rejected the second hypothesis, that participants reporting near falls retrospectively would report a fall prospectively. Although those results were not statistically significant, it was clinically relevant that retrospective near fallers would be twice as likely as their non-faller counterparts to report a fall prospectively.

The pass/fail result of balance tests identified that there were no significant differences in pass or fail between prospective near fallers and non-fallers for any of the tests under any of the conditions (Table 5.4 Balance test pass or fail status in prospective near fallers and non-fallers).

Table 5.4 Balance test pass or fail status in prospective near fallers and non-fallers

Test	Prospective near faller (pass/fail)	Prospective non-faller (pass/fail)	Chi-square	Significance p =	Effect size
Base Lun	41/22	45/14	1.84	0.18	0.12
Base SLS	57/6	53/6	0.01	0.91	-0.01
Base Tan	47/16	45/14	0.05	0.83	0.02
DT Lun	43/20	46/13	1.46	0.23	0.11
DT SLS	53/10	48/11	0.16	0.69	-0.04
DT Tan	42/21	45/14	0.37	0.24	0.11
SW Lun	44/19	47/12	1.55	0.21	0.11
SW SLS	52/11	52/7	0.76	0.38	0.08
SW Tan	41/22	42/17	0.52	0.47	0.07
Both Lun	45/18	44/15	0.15	0.70	0.04
Both SLS	57/6	47/12	2.83	0.09	-0.15
Both Tan	41/22	43/16	0.87	0.35	0.08
Final Lun	45/18	44/15	0.15	0.70	0.04
Final SLS	56/7	54/5	0.29	0.63	0.04
Final Tan	52/11	45/14	0.74	0.39	-0.08

Key Base baseline; Both distraction with fatigue; DT distracted condition; SW fatigued condition (incremental shuttle walk test); Lun lunge; SLS single leg stance; Tan tandem steps

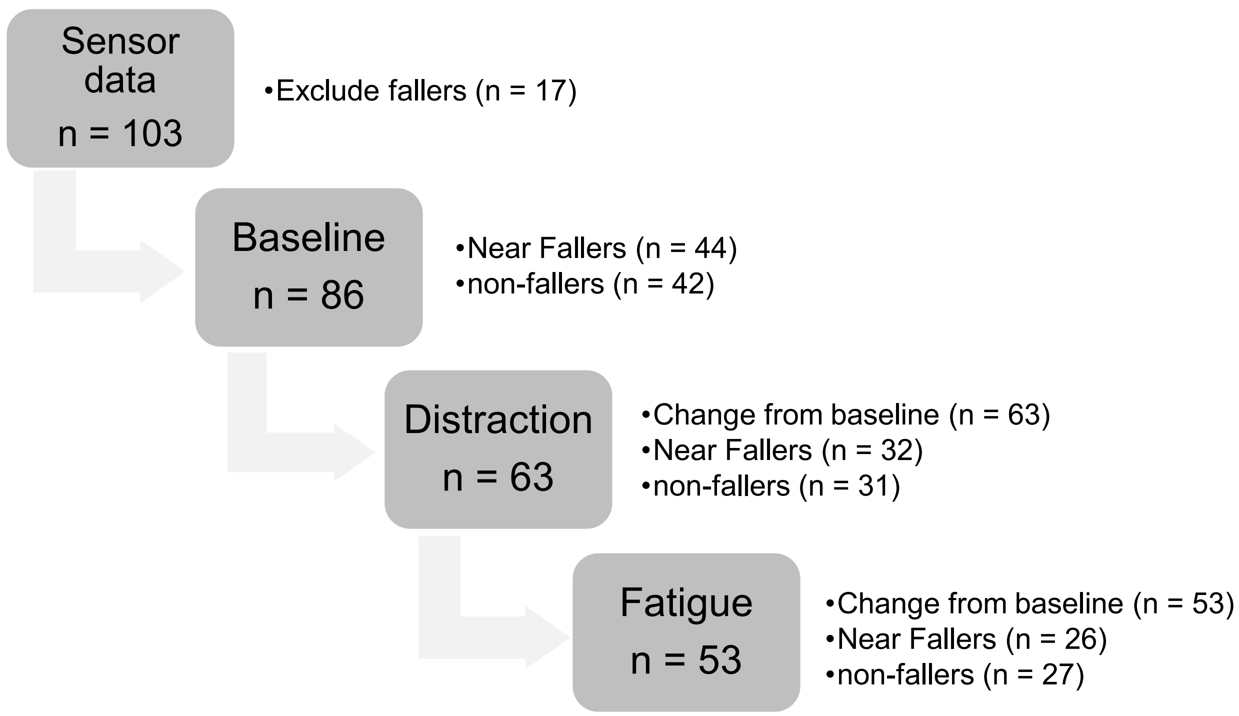
Chapter Summary

This chapter explored the contributing factors to near falls in midlife and young-older adults. Retrospectively reported near fallers were significantly more likely to report poorer general health and vitality, dizziness, poor vision, and fail the fatigued single leg stance. One quarter of the retrospective near fallers (10 of 38, 26.3%) went on to become fallers – clinically but not significantly relevant. Data on the prospectively reported near fallers was considered more consistent and provided new information not previously investigated. The prospectively reported near fallers and non-fallers were similar in all demographics, showing no significant differences for age, gender, self-reported biopsychosocial parameters, or balance test results. The similarity between near fallers and non-fallers required further investigation, so the next chapter explores the sway measures supplied by the sensors. These data provide more detailed and comprehensive information on the sway variables in the prospectively reported near faller and non-faller groups.

CHAPTER 6 SWAY RESULTS

The previous chapter investigated the clinical balance test results with relation to the contributing factors for near falls. This chapter reports the angular velocity, acceleration vector magnitude and RMS acceleration from the sensors. It provides insights into differences between prospectively reported near fallers and non-fallers and the impact of fatigue and distraction. Due to sensor dropout and problems with Bluetooth connection in the community venues, some sensor data were lost and data from 17 fallers were excluded from sway analyses - see sensor analyses flowchart figure 6.1. All sway results compare prospectively reported near fallers with prospectively reported non-fallers.

Figure 6.1 Sway analyses



Baseline

Whole Group Baseline Sway

To provide an understanding of the movement patterns in the lunge, single leg stance and tandem steps for the whole group, the raw data for each movement were presented (Table 6.1 Baseline sway differences between tests). Subsequently, Friedman's Tests were applied to the acceleration vector magnitude of lunge, single leg stance and tandem to gain an understanding of the whole movement magnitude between tests. Following this, the raw data for mediolateral, anteroposterior, and vertical acceleration, and pitch, roll and yaw angular velocities in each balance test were presented (Table 6.1). Results were depicted as Friedman test z score with Bonferroni-adjusted significance for multiple tests. The acceleration vector magnitude of sway significantly increased from single leg stance (Md 0.13 m/s^2) to tandem steps (Md 0.15 m/s^2) to lunge (Md 0.17 m/s^2) ($X^2_{(2)} = 55.79$, $p < 0.01$). Mediolateral sway was significantly different between the three balance tests ($X^2_{(2)} = 49.55$, $p < 0.01$) with the slowest acceleration in single leg stance (Md 0.11 m/s^2), followed by tandem steps (Md 0.13 m/s^2) and the fastest in lunge (Md 0.14 m/s^2). This pattern of significantly increased vertical acceleration also occurred from single leg stance to tandem to lunge ($X^2_{(2)} = 59.83$, $p < 0.01$). The anteroposterior sway acceleration was significantly slower in single leg stance (Md 0.03 m/s^2) than lunge (Md 0.05 m/s^2) and tandem steps (Md 0.04 m/s^2). The angular velocities produced similar patterns, significantly increasing pitch, roll and yaw velocities from single leg stance to tandem steps to lunge (Table 6.1). There was also a significant increase in angular velocity from single leg stance to tandem and lunge in the roll plane, indicating more lateral flexion at the trunk during the dynamic tests.

Table 6.1 Whole group baseline sway

Whole group raw measures of baseline sway (median, IQR)				
Whole group (axis)		**Lunge**	**SLS**	**Tandem**
Acceleration (m/s^2)	Vector Magnitude	0.17 (0.15, 0.19)	0.13 (0.12, 0.15)	0.15 (0.13, 0.17)
	RMS ML (x)	0.14 (0.12, 0.16)	0.11 (0.11, 0.13)	0.13 (0.11, 0.14)
	RMS AP (y)	0.05 (0.04, 0.06)	0.03 (0.02, 0.04)	0.04 (0.03, 0.05)
	RMS V (z)	0.08 (0.06, 0.10)	0.05 (0.04, 0.07)	0.06 (0.05, 0.09)
Angular velocity (rad/s)	Pitch (x) AV	1.28 (0.87, 1.86)	0.64 (0.39, 1.18)	1.02 (0.61, 1.69)
	Roll (y) AV	1.11 (0.75, 1.85)	0.52 (0.33, 0.91)	0.77 (0.55, 0.10)
	Yaw (z) AV	0.71 (0.53, 0.98)	0.39 (0.23, 0.63)	0.64 (0.48, 0.86)

Sway differences between tests Friedman's test z score (significance), *effect size*					
Baseline n = 86		**Overall**	**Lun - SLS**	**SLS - Tan**	**Tan - Lun**
Acceleration	Vector magnitude	**55.79 (0.00)**	**7.43 (0.00) *0.67***	**-3.15 (0.00)** *0.29*	**4.28 (0.00) *0.39***
	RMS ML (x)	**49.55 (0.00)**	**6.98 (0.00) *0.63***	**-2.81 (0.02)** *0.25*	**4.17 (0.00) *0.38***
	RMS AP (y)	**52.04 (0.00)**	**6.94 (0.00) *0.63***	**-5.12 (0.00) *0.46***	1.82 (0.21) *0.17*
	RMS V (z)	**59.83 (0.00)**	**7.70 (0.00) *0.70***	**-3.56 (0.00) *0.32***	**4.13 (0.00) *0.37***
Gyroscope	AV Pitch (x)	**49.98 (0.00)**	**6.79 (0.00) *0.73***	**-5.11 (0.00) *0.55***	1.68 (0.28) *0.15*
	AV Roll (y)	**43.28 (0.00)**	**6.56 (0.00) *0.71***	**-3.74 (0.00) *0.40***	**2.82 (0.01) *0.30***
	AV Yaw (z)	**35.27 (0.00)**	**5.45 (0.00) *0.59***	**-4.73 (0.00) *0.51***	0.72 (1.00) *0.07*

Key: AP anteroposterior; IQR interquartile range; Lun lunge; ML mediolateral; RMS root mean square; SLS single leg stance; Tan tandem steps; V vertical

Baseline Comparison of Near Fallers and Non-fallers

Comparison between near fallers (n = 44) and non-fallers (n = 42) in acceleration and angular velocity showed no significant differences for any baseline balance test (see Table 6.2. Near faller and non-faller baseline sway raw data).

Table 6.2 Near faller and non-faller baseline sway raw data

Baseline sway raw data (Median, IQR)							
Baseline Near Fallers n = 44; non-fallers n = 42		**Lunge**		**SLS**		**Tandem**	
		Near Fallers	non-fallers	Near Fallers	non-fallers	Near Fallers	non-fallers
Accelerometer Raw Data (m/s^2)	VM_a	0.17 (0.15, 0.19)	0.17 (0.15, 0.19)	0.13 (0.12, 0.15)	0.13 (0.12, 0.15)	0.14 (0.13, 0.17)	0.15 (0.14, 0.17)
	RMS ML (x)	0.14 (0.12, 0.16)	0.14 (0.13, 0.16)	0.11 (0.11, 0.13)	0.11 (0.11, 0.13)	0.13 (0.11, 0.14)	0.13 (0.11, 0.14)
	RMS AP (y)	0.05 (0.04, 0.06)	0.04 (0.03, 0.07)	0.03 (0.02, 0.04)	0.03 (0.02, 0.04)	0.04 (0.03, 0.05)	0.04 (0.03, 0.06)
	RMS V (z)	0.09 (0.06, 0.10)	0.08 (0.06, 0.10)	0.06 (0.04, 0.07)	0.05 (0.04, 0.07)	0.06 (0.05, 0.09)	0.07 (0.05, 0.08)
Gyroscope Raw Data (rad/s)	AV Pitch (x)	1.15 (0.76, 1.91)	1.30 (1.02, 1.88)	0.55 (0.37, 1.33)	0.74 (0.42, 1.16)	0.90 (0.63, 1.41)	1.11 (0.60, 1.83)
	AV Roll (y)	1.09 (0.68, 1.65)	1.16 (0.75, 2.14)	0.48 (0.33, 0.89)	0.32 (0.56, 1.02)	0.81 (0.54, 1.23)	0.77 (0.55, 1.05)
	AV Yaw (z)	0.70 (0.45, 1.07)	0.73 (0.56, .095)	0.34 (0.22, 0.79)	0.42 (0.23, 0.55)	0.61 (0.47, 0.88)	0.66 (0.51, 0.86)

Key: AP anteroposterior; AV angular velocity; ISR interquartile range; ML mediolateral; m/s^2 metres per second squared; rad/s radians per second; RMS root mean squared; V vertical; VM_a vector magnitude of acceleration.

The differences between the near fallers and non-fallers were analysed with Mann Whitney U tests (Table 6.3 Baseline sway differences between near fallers and non-fallers).

Table 6.3 Baseline sway differences between near fallers and non-fallers

Mann Whitney U test z score (sig)				
		Lunge	**SLS**	**Tandem**
Accelerometer	Vector Magnitude	-0.06 (0.95)	-0.35 (0.72)	-0.88 (0.38)
	RMS ML (x)	0.17 (0.86)	-0.18 (0.86)	-0.59 (0.55)
	RMS AP (y)	-0.16 (0.87)	-0.70 (0.48)	-1.23 (0.22)
	RMS V (z)	0.61 (0.54)	-0.12 (0.91)	0.17 (0.87)
Gyroscope	AV Pitch (x)	-0.88 (0.38)	-0.37 (0.71)	-1.15 (0.25)
	AV Roll (y)	-1.16 (0.25)	-0.29 (0.77)	0.47 (0.64)
	AV Yaw (z)	0.37 (0.71)	-0.31 (0.76)	-0.48 (0.63)

Key: AP anteroposterior; AV angular velocity; ML mediolateral; RMS root mean squared; V vertical.

These accelerometry and gyroscope results indicated that, compared to the non-fallers, the near fallers swayed in similar direction, speed, and angle during all baseline tests, and there were no significant differences between the two groups This rejects the hypothesis that near fallers would have significantly increased sway compared to non-fallers during the balance tests.

Once the baseline differences were understood, the effect of distraction on sway was investigated.

Distraction

Whole Group Distraction Compared to Baseline

During distraction, sway increased compared to baseline testing as identified by the median and interquartile ranges of the raw data (Table 6.4 Whole group distracted sway). Across the whole cohort, Wilcoxon Signed Rank Tests analysed the changes in sway from baseline to distraction.

Table 6.4 Whole group distracted sway

Whole group raw measures of distracted sway (median, IQR) (n = 63)				
Whole group		**Lunge**	**SLS**	**Tandem**
Acceleration (m/s^2)	VM_a	0.19 (0.17, 0.20)	0.15 (0.14, 0.18)	0.17 (0.15, 0.20)
	RMS ML (x)	0.15 (0.14, 0.16)	0.13 (0.12, 0.15)	0.14 (0.13, 0.16)
	RMS AP (y)	0.05 (0.04, 0.06)	0.04 (0.02, 0.06)	0.04 (0.03, 0.05)
	RMS V (z)	0.09 (0.08, 0.11)	0.07 (0.05, 0.09)	0.08 (0.06, 0.10)
Angular velocity (rad/s)	Pitch (x) AV	1.12 (0.80, 1.57)	0.73 (0.46, 1.31)	0.81 (0.60, 1.23)
	Roll (y) AV	0.97 (0.64, 1.40)	0.56 (0.39, 0.86)	0.71 (0.51, 1.08)
	Yaw (z) AV	0.68 (0.46, 0.91)	0.43 (0.27, 0.63)	0.63 (0.43, 0.87)
Whole group change baseline to distraction (n = 63) Wilcoxon Signed Rank Tests z score (sig) *effect size*				
Acceleration	VM_a	**3.74 (0.00) *0.47***	**4.27 (0.00) *0.54***	**4.55 (0.00) *0.57***
	RMS ML (x)	**3.34 (0.00) *0.42***	**4.96 (0.00) *0.62***	**4.53 (0.00) *0.57***
	RMS AP (y)	2.88 (0.77)	**3.31 (0.00) *0.41***	1.51 (0.13)
	RMS V (z)	**3.43 (0.00) *0.43***	**3.79 (0.00) *0.48***	**3.66 (0.00) *0.46***
Angular velocity	AV Pitch (x)	-1.67 (0.10)	1.30 (0.19)	-0.06 (0.95)
	AV Roll (y)	**2.60 (0.01) *0.33***	-0.13 (0.90)	-1.88 (0.06)
	AV Yaw (z)	-1.24 (0.22)	0.72 (0.47)	-0.56 (0.58)

Key: AP anteroposterior; AV angular velocity; IQR interquartile range; ML mediolateral; m/s^2 metres per second squared; rad/s radians per second; RMS root mean squared; V vertical; VM_a vector magnitude acceleration.

During distracted lunge, the whole cohort demonstrated a significant increase in sway acceleration, measured by vector magnitude acceleration ($z_{(63)} = 3.74$, $p < 0.01$, $r = 0.47$), compared to baseline. Directionally, there was a significant increase in sway acceleration mediolaterally ($z_{(63)} = 3.34$, $p < 0.01$, $r = 0.42$) and vertically ($z_{(63)} = 3.43$, $p < 0.01$, $r = 0.43$). There was an associated significant increase in the roll plane angular velocity ($z_{(63)} = 2.60$, $p = 0.01$, $r = 0.33$). These combine to describe an increased movement overall, with more lateral and vertical sway, more roll during distracted lunge compared to baseline lunge.

In the distracted single leg stance, the acceleration vector magnitude ($z_{(63)}$ =4.27, $p < 0.01$, $r = 0.54$) was significantly increased, indicating larger overall movement of the centre of mass during distracted single leg stance compared to baseline single leg stance. The rotation around the axes did not significantly change during distracted compared to baseline single leg stance. However, the directional acceleration along each of the three axes increased significantly during distracted single leg stance (mediolateral $z_{(63)} = 4.96$, $p < 0.01$, $r = 0.63$; anteroposterior $z_{(63)} = 3.31$, $p < 0.01$, $r = 0.42$; vertical $z_{(63)} = 3.79$, $p < 0.01$, $r = 0.48$) describing higher directional movement of the centre of mass during distracted single leg stance.

During distracted tandem steps, there was an increase in the magnitude of sway acceleration ($z_{(63)} = 4.55$, $p < 0.01$, $r = 0.57$) which incorporated significant increases in mediolateral ($z_{(63)} = 4.53$, $p < 0.01$, $r = 0.57$) and vertical sway ($z_{(63)} = 3.66$, $p < 0.01$, $r = 0.46$). These results indicate increased lateral movement during the distracted tandem steps compared to baseline.

Distracted Conditions Comparison of Near Fallers and Non-fallers

The raw data for the distracted conditions are presented in Table 6.5 Near faller and non-faller distracted sway raw data .

Table 6.5 Near faller and non-faller distracted sway raw data

Distracted sway raw data (Median, IQR)								
Near Fallers n = 32; non-fallers n = 31		**Lunge**		**SLS**		**Tandem**		
		Near Fallers	non-fallers	Near Fallers	non-fallers	Near Fallers	non-fallers
Accelerometer Raw data (m/s^2)	Vector Magnitude	0.19 (0.17, 0.21)	0.18 (0.17, 0.20)	0.15 (0.15, 0.19)	0.15 (0.14, 0.17)	0.18 (0.15, 0.20)	0.17 (0.15, 0.19)
	RMS ML (x)	0.15 (0.14, 0.17)	0.15 (0.14, 0.16)	0.14 (0.12, 0.17)	0.13 (0.12, 0.14)	0.14 (0.13, 0.18)	0.14 (0.13, 0.16)
	RMS AP (y)	0.05 (0.04, 0.07)	0.05 (0.04, 0.06)	0.03 (0.02, 0.06)	0.04 (0.02, 0.06)	0.05 (0.03, 0.06)	0.04 (0.03, 0.05)
	RMS V (z)	0.10 (0.08, 0.11)	0.09 (0.08, 0.11)	0.07 (0.06, 0.08)	0.07 (0.05, 0.09)	0.08 (0.06, 0.10)	0.07 (0.06, 0.10)
Gyroscope Raw data (rad/s)	Pitch (x) AV (rad/s)	1.02 (0.77, 1.57)	1.14 (0.89, 1.58)	0.74 (0.46, 1.25)	0.71 (0.46, 1.43)	0.79 (0.57, 1.19)	0.84 (0.66, 1.66)
	Roll (y) AV (rad/s)	0.97 (0.68, 1.50)	0.96 (0.63, 1.26)	0.58 (0.40, 1.33)	0.55 (0.38, 0.69)	0.69 (0.49, 1.04)	0.73 (0.51, 1.17)
	Yaw (z) AV (rad/s)	0.61 (0.46, 0.90)	0.73 (0.46, 0.91)	0.39 (0.27, 0.58)	0.50 (0.25, 0.72)	0.54 (0.37, 0.77)	0.69 (0.51, 0.97)

Key: AP anteroposterior; AV angular velocity; IQR interquartile range; ML mediolateral; m/s^2 metres per second squared; rad/s radians per second; RMS root mean squared; V vertical.

The comparison of near fallers with non-fallers for distracted sway was conducted with Mann Whitney U tests (Table 6.6 Distracted sway differences between near fallers and non-fallers).

Table 6.6 Distracted sway differences between near fallers and non-fallers

Near fallers n = 32; non-fallers n = 31 Mann Whitney U test z score (sig) *effect size*				
Between groups		**Lunge**	**SLS**	**Tandem**
Acceleration	VM_a	1.23 (0.22)	1.25 (0.21)	0.13 (0.89)
	RMS ML (x)	0.41 (0.68)	1.85 (0.06)	-0.26 (0.80)
	RMS AP (y)	0.96 (0.34)	0.39 (0.69)	0.71 (0.48)
	RMS V (z)	1.08 (0.28)	-0.08 (0.94)	0.13 (0.89)
Angular Velocity	AV Pitch (x)	-0.51 (0.61)	-0.17 (0.87)	-1.33 (0.18)
	AV Roll (y)	0.50 (0.62)	0.88 (0.38)	-0.22 (0.83)
	AV Yaw (z)	-0.78 (0.44)	-0.53 (0.60)	**-2.10 (0.04)** *0.27*

Key: AP anteroposterior; AV angular velocity; ML mediolateral; m/s^2 metres per second squared; rad/s radians per second; RMS root mean squared; V vertical; VM_a vector magnitude acceleration.

During the distracted tandem steps, near fallers experienced significantly less trunk rotation, measured by yaw angular velocity, than the non-fallers ($z_{(63)} = -2.10$, $p = 0.04$, $r = 0.27$). However, there was only a small effect size. While there were also lower forward flexion and side flexion velocities for near fallers during distracted tandem, the results were not significant. There were no significant differences between near fallers and non-fallers in any of the sway measures for the distracted lunge and distracted single leg stance.

Once the results for distraction were understood, the differences in fatigued sway were investigated.

Fatigue

Whole Group Fatigue Compared to Baseline

Similar to the performance of balances tests while distracted, fatigued sway compared to baseline resulted in a significant increase in sway acceleration vector magnitude for each of the three tests: lunge ($z = 5.88$, $p < 0.01$, $r = 0.81$), single leg stance ($z = 4.82$, $p < 0.001$, $r = 0.66$) and tandem steps ($z = 5.67$, $p < 0.001$, $r = 0.78$) (Table 6.7 Whole group fatigued sway). The effect sizes for these increases were all large ($r > 0.5$), indicating a strong relationship between fatigue and increased sway magnitude.

Table 6.7 Whole group fatigued sway

Whole group raw measures of fatigued sway (median, IQR) **(n = 53)**				
Whole group		**Lunge**	**SLS**	**Tandem**
Acceleration (m/s^2)	VM_a	0.22 (0.20, 0.26)	0.17 (0.15, 0.19)	0.19 (0.18, 0.23)
	RMS ML (x)	0.18 (0.16, 0.19)	0.14 (0.13, 0.16)	0.16 (0.15, 0.19)
	RMS AP (y)	0.06 (0.05, 0.08)	0.03 (0.02, 0.05)	0.06 (0.04, 0.08)
	RMS V (z)	0.12 (0.10, 0.14)	0.07 (0.05, 0.09)	0.09 (0.07, 0.10)
Angular velocity (rad/s)	Pitch (x) AV	1.28 (0.92, 1.80)	0.79 (0.46, 1.13)	1.05 (0.72, 1.66)
	Roll (y) AV	1.13 (0.75, 1.75)	0.61 (0.41, 1.00)	0.83 (0.58, 1.35)
	Yaw (z) AV	0.76 (0.56, 0.96)	0.46 (0.31, 0.61)	0.73 (0.55, 0.93)
Whole group change baseline to fatigue (n = 53) Wilcoxon Signed Rank Tests z score (sig), *effect size*				
Accelerometer	VM_a	**5.88 (0.00) *0.81***	**4.82 (0.00) *0.59***	**5.67 (0.00) *0.78***
	RMS ML (x)	**5.67 (0.00) *0.78***	**5.38 (0.00) *0.74***	**5.79 (0.00) *0.80***
	RMS AP (y)	**4.56 (0.00) *0.63***	**2.62 (0.01) *0.36***	**4.52 (0.00) *0.62***
	RMS V (z)	**6.20 (0.00) *0.85***	**3.34 (0.00) *0.46***	**4.94 (0.00) *0.68***
Angular velocity	AV Pitch (x)	0.77 (0.44)	**2.01 (0.05)** *0.28*	**2.21 (0.03) *0.30***
	AV Roll (y)	0.15 (0.88)	1.25 (0.21)	0.26 (0.79)
	AV Yaw (z)	1.59 (0.11)	1.66 (0.10)	**2.08 (0.04)** *0.29*

Key: AP anteroposterior; AV angular velocity; ML mediolateral; m/s^2 metres per second squared; rad/s radians per second; RMS root mean squared; V vertical; VM_a vector magnitude acceleration.

Across the cohort during the lunge each of mediolateral, anteroposterior, and vertical direction accelerations were significantly increased compared to baseline, but there was no significant difference in the pitch, roll or yaw planes of rotation. During the fatigued single leg stance, the whole cohort demonstrated a significant increase in mediolateral, anteroposterior, and vertical accelerations compared to baseline. The fatigued single leg stance also produced a significant increase in pitch angular velocity compared to baseline single leg stance. During the fatigued tandem steps, there were significantly increased sway accelerations across the cohort in the mediolateral, anteroposterior, and vertical planes, as well as the pitch and yaw angular velocities compared to baseline tandem (Table 6.7 Whole group fatigued sway).

Fatigued Conditions Comparison of Near Fallers to Non-fallers

The raw data for the differences in fatigued sway between near fallers and non-fallers is presented in Table 6.8 Fatigued sway raw data for near fallers and non-fallers.

Table 6.8 Near faller and non-faller fatigued sway raw data

Fatigued sway raw data (Median, IQR) Near Fallers n = 26; non-fallers n = 27							
Between groups		**Lunge**		**SLS**		**Tandem**	
		Near fallers	non-fallers	Near fallers	non-fallers	Near fallers	non-fallers
Accelerometer Raw data (m/s^2)	Acceleration Magnitude	0.22 (0.20, 0.26)	0.22 (0.19, 0.25)	0.16 (0.15, 0.19)	0.17 (0.15, 0.18)	0.19 (0.18, 0.22)	0.20 (0.18, 0.24)
	RMS ML (x)	0.18 (0.15, 0.19)	0.17 (0.16, 0.20)	0.14 (0.13, 0.16)	0.14 (0.13, 0.16)	0.17 (0.15, 0.19)	0.16 (0.15, 0.20)
	RMS AP (y)	0.07 (0.04, 0.08)	0.06 (0.05, 0.08)	0.03 (0.02, 0.05)	0.03 (0.02, 0.06)	0.05 (0.04, 0.08)	0.06 (0.04, 0.08)
	RMS V (z)	0.12 (0.10, 0.14)	0.12 (0.09, 0.14)	0.07 (0.05, 0.08)	0.07 (0.05, 0.10)	0.08 (0.07, 0.10)	0.09 (0.08, 0.11)
Gyroscope Raw data (rad/s)	Pitch (x) AV (rad/s)	1.31 (0.52, 1.78)	1.27 (0.91, 1.85)	0.70 (0.37, 1.08)	0.83 (0.58, 1.25)	1.05 (0.73, 1.43)	1.06 (0.69, 1.79)
	Roll (y) AV (rad/s)	1.13 (0.79, 1.89)	1.13 (0.67, 1.74)	0.57 (0.37, 1.06)	0.64 (0.47, 0.97)	0.81 (0.60, 1.34)	0.82 (0.51, 1.41)
	Yaw (z) AV (rad/s)	0.69 (0.52, 0.98)	0.82 (0.63, 0.98)	0.35 (0.25, 0.60)	0.53 (0.40, 0.80)	0.75 (0.54, 1.00)	0.69 (0.55, 0.93)

Key: AP anteroposterior; AV angular velocity; ML mediolateral; m/s^2 metres per second squared; rad/s radians per second; RMS root mean squared; V vertical.

Subsequently, the differences between the near fallers and non-fallers were analysed with Mann Whitney U tests and the results are displaced in Table 6.9 Fatigued sway differences between near fallers and non-fallers.

Table 6.9 Fatigued sway differences between near fallers and non-fallers

Between group differences in fatigued tests Mann Whitney U test z score (sig) *effect size*				
Between groups		**Lunge**	**SLS**	**Tandem**
Accelerometer	VM_a	0.42 (0.67)	-0.59 (0.56)	-0.66 (0.51)
	RMS ML (x)	-0.26 (0.80)	-0.80 (0.43)	0.03 (0.98)
	RMS AP (y)	-0.19 (0.85)	-0.09 (0.93)	-0.63 (0.53)
	RMS V (z)	0.84 (0.40)	-0.16 (0.88)	-0.16 (0.12)
Gyroscope	AV Pitch (x)	0.15 (0.88)	-1.36 (0.17)	-0.18 (0.86)
	AV Roll (y)	0.76 (0.45)	-0.66 (0.51)	0.16 (0.87)
	AV Yaw (z)	-0.53 (0.59)	**-2.23 (0.03) *0.31***	0.14 (0.89)

Key: AP anteroposterior; AV angular velocity; ML mediolateral; m/s^2 metres per second squared; rad/s radians per second; RMS root mean squared; V vertical; VM_a vector magnitude acceleration.

Near fallers had significantly less trunk rotation, indicated by yaw velocity, than the non-fallers during the single leg stance ($z_{(53)} = -2.23$, $p = 0.03$, $r = 0.31$). There was a moderate effect size for this result. All other results showed no significant differences between near fallers and non-fallers during the fatigued balance tests.

Chapter summary

This chapter explored the research questions 3, 4 and 5 on the sway differences between near fallers and non-fallers at baseline, the sway differences due to distraction and fatigue, and the effect of the distraction and fatigue on the near fallers compared to non-fallers. Across the whole group at baseline there was significantly increased sway acceleration vector magnitude, RMS acceleration along each axis, and angular velocity across all planes from single leg stance to tandem steps to lunge. This was somewhat expected due to the progression from static position in the single leg stance, to dynamic movement forwards with tandem steps, and the additional vertical component in the lunge. However, when sway in the near fallers was compared to non-fallers at baseline, there were no significant differences in sway acceleration vector magnitude, RMS acceleration on any axis, or angular velocity in any plane for any of the balance tests. This rejects the hypothesis that near fallers would demonstrate significantly increased sway compared to non-fallers at baseline testing.

In distracted sway across the cohort there was a significant increase in sway acceleration vector magnitude, mediolateral and vertical RMS acceleration in lunge, single leg stance and tandem steps compared to baseline. During the lunge there was an additional increased lateral flexion of the trunk during distraction compared to baseline. Compared to non-fallers, near fallers produced significantly less trunk rotation during distracted tandem steps. These results reject the hypothesis that near fallers would have significantly increased sway compared to non-fallers for the distracted tests.

In the fatigued conditions the whole group produced significantly increased sway acceleration vector magnitude and increased RMS acceleration during all three tests compared to baseline. Similarly, across the whole cohort there were significant increases in trunk and hip flexion during fatigued single leg stance and tandem steps, and also increased trunk rotation in fatigued tandem steps, compared to baseline, as demonstrated by the angular velocity results. Near fallers demonstrated significantly less trunk rotation than non-fallers during single leg stance. This rejects the hypothesis that near fallers would have significantly increased sway compared to non-fallers during fatigued testing. The results on the differences between near fallers and non-fallers during baseline, distracted and fatigued tests reject all the hypotheses that near fallers would demonstrate significantly increased sway compared to non-fallers. In contrast, near fallers produced significantly less trunk rotation compared to non-fallers during select tests.

Once the sway measures from balance tests comparing the near fallers and non-fallers were understood, it was necessary to investigate the predictive ability of sway to identify near fallers.

CHAPTER 7 PREDICTIVE ABILITY OF SWAY TO IDENTIFY NEAR FALLERS

A forward stepwise logistic regression approach was used to explore the influence of potential predictors on the outcome of near falls. Predictor variables were grouped as previously described (Chapter 4 Methods and Protocol, p. 44):

- Demographics – self-reported descriptors of the participant cohort:
    - Age, gender, living status, work status
- Biometrics – measures of participants' physical characteristics:
    - Resting heart rate, body mass index, hearing, vision, dizziness
- Experiential - self reports of emotion and self-perceived function:
    - Fear of falling, SF36 general health, SF36 physical function , SF36 vitality
- Baseline sway data for the lunge, single leg stance and tandem steps:
    - Vector magnitude acceleration, RMS acceleration (mediolateral, anteroposterior, and vertical), angular velocity (pitch, roll and yaw)

The number of participants with sway data was n = 86 for baseline, n = 63 for distraction and n = 53 for fatigue. As the minimum sample size was based on the formula outlined in Chapter 4 (p. 46) ($n > 50 + 8m$), or n > 66 for two predictor variables, the distraction and fatigue sway variables were not included in the prediction model. Variables were assessed for multicollinearity by correlation matrices (Table 7.1 – 7.3 Correlation analyses).

Correlation analyses

Table 7.1 Correlation analysis demographic and biometric variables

A: demographic and biometric variable correlations							
Spearman's rho correlation n = 122		Age	Resting HR	BMI	Gen Hlth	Vitality	Phys Fun
Age	CC	1.00					
Resting heart rate	CC	.008	1.00				
Body Mass Index	CC	.114	-.003	1.00			
General Health	CC	.165	.009	-.124	1.00		
Vitality	CC	.154	.079	-.215*	**.639****	1.00	
Physical Function	CC	-.289**	-.100	-.372**	.404**	.483**	1.00

Key: BMI body mass index; CC correlation coefficient ; HR heart rate; Gen Hlth SF36 general health; Phys Fun SF36 physical function

Table 7.2 Correlation analysis RMS and vector magnitude acceleration

B: Baseline vector magnitude acceleration and RMS acceleration correlations													
Spearman's rho correlation n = 86		VM_a L	VM_a S	VM_a T	RMS_a L ML	RMS_a L AP	RMS_a L V	RMS_a S ML	RMS_a S AP	RMS_a S V	RMS_a T ML	RMS_a T AP	RMS_a T V
VM_a L	CC	1.000											
VM_a S	CC	.382**	1.000										
VM_a T	CC	.339**	.329**	1.000									
RMS_a L ML	CC	**.878****	.271*	.265*	1.000								
RMS_a L AP	CC	**.730****	.302**	.223*	**.554****	1.000							
RMS_a L V	CC	**.603****	.362**	.273*	.254*	.425**	1.000						
RMS_a S ML	CC	.313**	**.775****	.192	.415**	.158	.050	1.000					
RMS_a S AP	CC	.230*	**.673****	.191	.163	.409**	.173	**.510****	1.000				
RMS_a S V	CC	.185	**.661****	.313**	-.092	.168	**.592****	.181	.437**	1.000			
RMS_a T ML	CC	.312**	.184	**.847****	.396**	.157	.064	.252*	.054	-.010	1.000		
RMS_a T AP	CC	.331**	.261*	**.664****	.242*	.377**	.141	.054	.362**	.236*	.402**	1.000	
RMS_a T V	CC	.035	.221*	**.681****	-.188	.080	.386**	-.136	.137	**.632****	.311**	.470**	1.000

Table 7.3 Correlation analysis RMS acceleration and angular velocity

C: Baseline RMS acceleration and angular velocity correlations																			
Spearman's rho correlation (n = 86)		RMS$_a$ L ML	RMS$_a$ L AP	RMS$_a$ L V	RMS$_a$ S ML	RMS$_a$ S AP	RMS$_a$ S V	RMS$_a$ T ML	RMS$_a$ T AP	RMS$_a$ T V	AV L pitch	AV L roll	AV L yaw	AV S pitch	AV S roll	AV S yaw	AV T pitch	AV T roll	AV T yaw
RMS$_a$ L ML	CC	1.00																	
RMS$_a$ L AP	CC	**.554****	1.00																
RMS$_a$ L V	CC	.254*	.425**	1.00															
RMS$_a$ S ML	CC	.415**	.158	.050	1.00														
RMS$_a$ S AP	CC	.163	.409**	.173	**.510****	1.00													
RMS$_a$ S V	CC	-.092	.168	**.592****	.181	.437**	1.00												
RMS$_a$ T ML	CC	.396**	.157	.064	.252*	.054	-.010	1.00											
RMS$_a$ T AP	CC	.242*	.377**	.141	.054	.362**	.236*	.402**	1.00										
RMS$_a$ T V	CC	-.188	.080	.386**	-.136	.137	**.632****	.311**	.470**	1.00									
AV L pitch	CC	.241*	.292**	.145	.139	.126	.057	.306**	.424**	.223*	1.00								
AV L roll	CC	.243*	.370**	.404**	.043	.092	.203	.211	.262*	.152	.317**	1.00							
AV L yaw	CC	.219*	.410**	.442**	.121	.283**	.282**	.348**	.420**	.366**	**.529****	**.551****	1.00						
AV S pitch	CC	.371**	.293**	-.054	.274*	.264*	.046	.209	.335**	-.045	.330**	-.002	.191	1.00					
AV S roll	CC	.229*	.204	.120	.334**	.257*	.201	.312**	.316**	.095	.315**	.083	.312**	**.713****	1.00				

AV S yaw	CC	.313**	.298**	.094	.310**	.238*	.095	.249*	.251*	.063	.230*	.094	.340**	**.822****	**.751****	1.00			
AV T pitch	CC	.234*	.100	-.065	.193	.281**	.067	.255*	.506**	.116	.463**	.055	.284**	**.509****	.335**	.369**	1.00		
AV T roll	CC	.288**	.165	.351**	.271*	.160	.324**	.341**	.302**	.254*	.298**	.330**	.480**	.287**	.454**	.388**	.477**	1.00	
AV T yaw	CC	.274*	.102	.225*	.153	.077	.124	.388**	.313**	.200	.311**	.328**	**.532****	.368**	.345**	.412**	**.651****	**.609****	1.00

Key: AP anteroposterior; AV angular velocity; CC Spearman's correlation coefficient; L lunge; ML mediolateral; n number in sample; RMS$_a$ root mean square acceleration; S single leg stance; T tandem; V vertical; VM$_a$ vector magnitude acceleration; ** correlation is significant at the 0.01 level (2-tailed); * correlation is significant at the 0.05 level (2-tailed)

There was a strong correlation between the experiential SF36 general health and SF36 vitality ($\rho = 0.64$, $p < 0.01$) which was not considered strong enough to confound the model. Very strong Spearman correlation coefficients of $r \geq 0.8$ (shown in **red** in correlation tables) were evident between

- Lunge vector magnitude acceleration and lunge mediolateral RMS acceleration
- Tandem vector magnitude acceleration and tandem mediolateral RMS acceleration
- Single leg stance angular velocity pitch and single leg stance angular velocity yaw

As the vector magnitude acceleration was a composite of the three axes RMS acceleration, the vector magnitude variables were excluded on the basis of singularity. Given that none of the baseline tests RMS acceleration or angular velocity variables had shown any significant differences between near fallers and non-fallers (Chapter 6), they were both included for univariate analysis to inform best choice of sway variable for the model.

The next step in the regression analysis was to conduct univariate analyses to identify the relationship between the each of the demographic, biometric, experiential and sway RMS acceleration variables with prospectively reported near falls. Each variable was entered separately (Table 7.4 Demographic and biometric univariate analysis for near fall outcome; Table 7.5 Experiential univariate analysis for near fall outcome; Table 7.6 Sway RMS acceleration univariate analysis for near fall outcome; Table 7.7 Sway angular velocity univariate analysis for near fall outcome).

Univariate analyses for near falls outcome

Table 7.4 Demographic and biometric univariate analysis for near fall outcome

Variables in the Equation								
Demographics	B	SE	Wald	df	Sig.	OR	95% CI for OR	
							Lower	Upper
Age	-0.01	0.02	0.10	1	0.76	0.994	0.958	1.03
Gender	0.18	0.43	0.17	1	0.68	1.193	0.518	2.75
Living Status	-0.57	0.41	1.91	1	0.17	0.565	0.252	1.27
Work Status	-0.12	0.37	0.10	1	0.75	0.889	0.435	1.82
Variables in the Equation								
Biometrics	B	SE	Wald	df	Sig.	OR	95% CI for OR	
							Lower	Upper
Resting HR	0.03	0.02	1.98	1	0.16	1.025	0.990	1.06
BMI	0.004	0.04	0.01	1	0.92	1.004	0.935	1.08
Dizziness	0.64	0.41	2.44	1	0.12	1.899	0.849	4.25
Vision	0.33	0.41	0.64	1	0.43	1.388	0.620	3.11
Hearing	-0.08	0.49	0.03	1	0.87	0.925	0.354	2.41

Key: B unstandardised regression weight; BMI body mass index; CI confidence interval; df degrees of freedom; HR heart rate; OR odds ratio; SE standard error of the estimate; Sig significance; Wald chi square test.

Table 7.5 Experiential univariate analysis for near fall outcome

Variables in the Equation								
Experiential	B	SE	Wald	df	Sig.	OR	95% CI for OR	
							Lower	Upper
Fear of falling	0.35	0.39	0.82	1	0.37	1.421	0.663	3.04
General Health	-0.01	0.01	0.28	1	0.60	0.995	0.976	1.01
Vitality	-0.01	0.01	0.22	1	0.64	0.995	0.974	1.02
Physical Function	0.02	0.01	**4.14**	**1**	**0.04**	**1.024**	**1.001**	**1.05**

Key: B unstandardised regression weight; CI confidence interval; df degrees of freedom; OR odds ratio; SE standard error of the estimate; Sig significance; Wald chi square test.

Table 7.6 Sway RMS acceleration univariate analysis for near fall outcome

Variables in the Equation								
Sway RMS acceleration	B	SE	Wald	df	Sig.	OR	95% CI for OR	
							Lower	Upper
RMS_a Lun ML	-0.64	5.24	0.02	1	0.90	0.53	0.00	15373.6
RMS_a Lun AP	0.82	7.77	0.01	1	0.92	2.27	0.00	9411662.1
RMS_a Lun V	5.04	6.47	0.61	1	0.44	153.65	0.00	49713554.3
RMS_a SLS ML	-5.16	6.10	0.72	1	0.40	0.01	0.00	893.2
RMS_a SLS AP	2.15	7.28	0.09	1	0.77	8.58	0.00	13412130.8
RMS_a SLS V	3.88	5.78	0.45	1	0.50	48.28	0.00	3980339.7
RMS_a Tan ML	-12.26	9.48	1.67	1	0.20	0.00	0.00	558.8
RMS_a Tan AP	-14.27	9.80	2.12	1	0.15	0.00	0.00	137.6
RMS_a Tan V	5.60	7.80	0.52	1	0.47	271.35	0.00	1173359248.4

Key: AP anteroposterior; B unstandardised regression weight; CI confidence interval; df degrees of freedom; Lun lunge; ML mediolateral; OR odds ratio; Sig significance; RMS_a root mean square acceleration; SE standard error of the estimate; SLS single leg stance; Tan tandem steps; V vertical; Wald chi square test.

Table 7.7 Sway angular velocity univariate analysis for near fall outcome

Variables in the Equation								
Sway angular velocity	B	SE	Wald	df	Sig.	OR	95% CI for OR	
							Lower	Upper
AV Lun pitch	0.01	0.24	0.00	1	0.98	1.01	0.63	1.61
AV Lun roll	-0.44	0.29	2.34	1	0.13	0.65	0.37	1.13
AV Lun yaw	0.75	0.51	2.17	1	0.14	2.13	0.78	5.80
AV SLS pitch	0.09	0.26	0.12	1	0.73	1.09	0.66	1.80
AV SLS roll	0.10	0.43	0.05	1	0.82	1.10	0.47	2.57
AV SLS yaw	0.30	0.48	0.39	1	0.53	1.35	0.52	3.49
AV Tan pitch	-0.41	0.28	2.19	1	0.14	0.67	0.39	1.14
AV Tan roll	0.54	0.43	1.61	1	0.20	1.72	0.75	3.95
AV Tan yaw	0.28	0.57	0.25	1	0.62	1.33	0.44	4.04

Key: AV angular velocity; B unstandardised regression weight; CI confidence interval; df degrees of freedom; Lun lunge; OR odds ratio; Sig significance; SE standard error of the estimate; SLS single leg stance; Tan tandem steps; Wald chi square test.

The self-reported SF36 Physical Function score was the only predictor variable to have a significant Wald score (χ^2 (1, 122) = 8.77, $p < 0.01$) with confidence intervals (C.I.) not crossing 1.0 (OR 1.06, 95% C.I. 1.02 - 1.11). Independently, physical function correctly classified 58.2% of cases (increase from 51.6% without independent variables). The odds ratio was positive, demonstrating that higher physical function scores were predictive of near falls. As higher physical function scores indicated better physical status (see Appendix I, SF36 Physical Function survey questions), this meant that participants with more physical ability and function were correctly predicted to be the near fallers.

The only distracted sway variable to show a significant difference between near fallers and non-fallers was yaw angular velocity, i.e., trunk rotation, in distracted tandem steps (Chapter 6 distracted sway results, p. 73). Similarly, the only significant difference in fatigued testing between the two groups was yaw angular velocity during single leg stance (Chapter 6 fatigued sway results p. 76). While the univariate regression analyses for these two predictor variables suggested an influence on prediction (distraction χ^2 (1, 63) = 3.92, $p = 0.05$, OR 0.20 (95% CI 0.04 - 0.98); fatigue χ^2 (1, 53) = 4.44, $p = 0.04$, OR 0.08 (95% CI 0.01 - 0.84)) there were too few participants with distracted and fatigued sway data for them to be considered definitively for the model.

As no other demographic, biometric, experiential, sway acceleration or sway angular velocity variables had significant chi square results or odds ratios confidence intervals excluding 1.0, they were not retained for trial in the prediction model and the stepwise logistic regression modelling was aborted for the dependent variable of near fall. The prediction model for identifying near fallers was not considered useful. The hypothesis that sway would be an independent predictor of near falls was rejected.

Subsequently, as physical function was the only significant variable to predict near falls, the relationship between physical function and inertial sensors was queried, to inform community practice and future research. A new hypothesis was generated: hypothesis 7 sway outcomes are related to SF36 physical function variance. To explain the variance in physical function scores from the sway variables, a regression analysis was completed with physical function as the dependent outcome. As the physical function results were not normally distributed, the cases were ranked using Blom's formula to provide a more normal distribution for use in multiple regression (Pallant, 2020). Univariate analysis was conducted to identify the significant sway variables for the regression model. The sway variables' correlations and the variance (R^2) on predicting physical function outcome were assessed for suitability to add to the model (Field, 2018). Significant findings were added to the hierarchical regression model to inform which sway variables increased the variance in physical function outcome. The more proximal relationships between the sway variables and the physical function outcome were analysed to investigate if sway predicted the physical function (Table 7.8 Sway variables univariate analysis for physical function outcome).

Univariate analyses for physical function outcome

Table 7.8 Sway variables univariate analysis for physical function outcome

Variable	Correlation	R^2	Adjusted R^2	Significance
Lunge RMS_a mediolateral	0.02	0.00	-0.01	0.86
Lunge RMS_a anteroposterior	-0.01	0.00	-0.01	0.94
Lunge RMS_a vertical	-0.09	0.01	0.00	0.42
SLS RMS_a mediolateral	-0.02	0.00	-0.01	0.85
SLS RMS_a anteroposterior	-0.10	0.01	0.00	0.35
SLS RMS_a vertical	-0.13	0.02	0.01	0.23
Tandem RMS_a mediolateral	0.00	0.00	-0.01	0.99
Tandem RMS_a anteroposterior	-0.13	0.02	0.01	0.22
Tandem RMS_a vertical	-0.12	0.02	0.00	0.27
Lunge AV pitch	0.12	0.01	0.00	0.28
Lunge AV roll	0.04	0.00	-0.01	0.71
Lunge AV yaw	0.05	0.00	-0.01	0.67
SLS AV pitch	0.03	0.00	-0.01	0.81
SLS AV roll	0.05	0.00	-0.01	0.65
SLS AV yaw	-0.03	0.00	-0.01	0.76
Tandem AV pitch	0.13	0.02	0.00	0.25
Tandem AV roll	0.07	0.01	-0.01	0.52
Tandem AV yaw	0.20	0.04	0.03	0.06

Key: AV angular velocity; RMS_a root mean square acceleration; R^2 proportion of variance explained by independent variable; SLS single leg stance

There were no strong correlations between the baseline sway RMS acceleration or angular velocity with physical function scores. None of the individual sway RMS accelerations or angular velocity variables explained any significant variance in the model, so a hierarchical regression model was not attempted. Sway RMS acceleration and angular velocity were not considered independent predictors for the physical function outcome, thereby rejecting the hypothesis.

Chapter Summary

In conclusion, physical function outcome was the only significant predictor of near falls. There was no further significant variance by adding any sway variables, indicating that sway was not a significant, independent predictor of near falls in this cohort. The hypothesis of sway predicting near falls was also rejected. Given that the sway variables did not provide an independent prediction for near falls, the data were not explored to find cut-off points for sway.

Sway was not an independent predictor for near falls. The next chapter interprets the results from demographic and sway investigations in response to the research questions.

CHAPTER 8 DISCUSSION

This was the first study to investigate near fallers as a distinct cohort. Balance was measured in prospectively reported near fallers, and the contributing factors to near falls were investigated. Postural sway in near fallers and non-fallers was assessed by clinical tests in static and dynamic balance. Postural sway was measured by gyroscope angular velocity, and root mean square acceleration and vector magnitude acceleration. The impact of fatigue and distraction on the balance and sway outcomes were scrutinized. This chapter describes the new findings of near fallers in relation to the current literature and responds to the hypotheses posed through the thesis.

Demographics

The first hypothesis was that near fallers would be older, afraid of falling, report hearing and vision impairment, dizziness, poor general health, low vitality and poor physical function, and be underweight or obese. This is similar to how fallers are described in the literature. However, there were no significant associations between prospectively reported near falls and any of these variables, rejecting the hypothesis. Unlike falls that are more associated with older age (Lusardi et al., 2017), near falls were not significantly related to age in this cohort. This suggests that age itself does not have a direct influence on near falls and that it is likely related to other factors.

Ageing affected sensory systems (Escamilla-Martínez et al., 2021) were not significantly associated with this group of near fallers. While self-reported poor vision and dizziness were significantly associated with retrospective reports of near falls, the association was lost with the more reliable prospective reports of near falls. In this study self-report measures provided a broad personal perspective of these sensory balance control systems. Neither vision nor vestibular control of balance were objectively assessed through activities such as the Sensory Organisation Test (Karmali et al., 2021) or the Clinical Test of Sensory Interaction and Balance (Ojie & Saatchi, 2021), which would provide more concrete information in future studies. Self-reported hearing disability was not associated with near falls as it is with fallers (Horowitz et al., 2020). As mild hearing loss is prevalent in midlife Australian adults (Wang et al., 2019), future research using objective audiometry testing instead of relying on self-report may produce different results.

Fear of falling was reported as a binary response and showed no significant relationship with prospective reports of near falls. In an earlier study on near fallers (Srygley et al., 2009), there were significant associations with anxiety and depression, rather than fear of falling. Anxiety and depression have a negative effect on activity levels that negatively impact physical function (Painter et al., 2012). This pattern can become a cycle of fear avoidance, reducing physical function and further increasing the fear of falling (van Haastregt et al., 2008). More current research also suggests a link between past falls and fear of falls (Peeters et al., 2020). The cohort in this study only included non-fallers which may have been why there was no association with fear of falling in this project. Much of the literature regarding fear of falling investigates known fallers, older adults or after diagnosis of a neurological condition. In midlife adults, fear of falling has been investigated, but only in neurological conditions such as stroke (Sanchez-Sanchez et al., 2021) or Parkinson's Disease (Rutz & Benninger, 2020) which were exclusion criteria for this study. Other groups of midlife adults investigated for fear of falling have specific symptoms such as low blood pressure (de Souza et al., 2015), dizziness (Holle et al., 2015), or fibromyalgia (Chiaramonte et al., 2019). Fear of falling is considered a single component of confidence within the complexity of quality of life, physical function, and physical activity (Schoene et al., 2019). In this cohort of seemingly healthy adults the higher physical function and general health results would indicate more confidence than fear of falling during functional mobility. This confidence would encourage higher or more frequent activity levels, which subsequently could provide more frequent opportunities for slips, trips and missteps.

Body mass index categories of underweight and obese are significantly associated with falls (Ogliari et al., 2021a). Being underweight is an established indicator of pre-frailty or frailty (Fried et al., 2001), and a predictor of falls risk in older adults (Trevisan et al., 2019). However, underweight was not significantly associated with near falls. Obesity in midlife is significantly associated with falls risk in women (Karvonen-Gutierrez et al., 2020). Weight gain due to menopause in this age group can also negatively impact functional activity (Knight et al., 2021). However, there was no significant association between obesity and near falls in this cohort.

Based on the previous research using retrospective data (Baker et al., 2021a), near fallers were expected to fail the balance tests. There was a significant association for near fallers to fail the single leg stance under fatigued conditions, but only in the retrospectively grouped near fallers. The association did not persist with prospectively reported near falls. Previously reported near falls were significant enough to remember to report, suggesting that the recovery movement was sufficiently distinct to be memorable.

Near fallers were similar to non-fallers, as demonstrated by the lack of differences in the demographic reports. As none of the demographic or clinical test outcomes had a significant association with prospectively reported near falls, the first hypothesis was rejected. The demographic findings identify that prospectively reported near fallers do not present the same as retrospectively reported near fallers, nor the same characteristics expected of fallers. They do, however, present with good self-reported physical function and general health, healthy to slightly overweight weight to height ratio, and with confidence in their functional mobility.

Becoming fallers

The second hypothesis was that retrospectively reported near fallers would become fallers by the end of the study. There was no statistically significant association between retrospectively reported near falls and prospectively reported near falls or falls, rejecting the hypothesis. However, the odds ratio results, whilst not statistically significant, indicated a clinically relevant finding that near fallers were 2.2 times more likely to fall than their non-faller counterparts. The three-month prospective diary entry data collection timeframe in this study therefore aligns with the results from previous studies where older community-living adults were more likely to sustain a fall within three weeks (Ryan et al., 1993), or six months (Nagai et al., 2017) to a year (Srygley et al., 2009) of a near fall. The lack of statistically significant difference in this study cohort may be due more to the younger, midlife participants within this study, median age 64 years. The proportion of people reporting near falls from retrospective reporting (26%) was less than prospective reporting (43%), which supports and reinforces prospective data collection for more detailed information. The lack of significant findings for near fallers becoming fallers within the timeframe of the study placed new attention on the third hypothesis.

Baseline sway

The third hypothesis was that near fallers would demonstrate increased baseline sway during the balance tasks, compared to non-fallers. Physical performance reduces as the complexity of a physical task increases (Brustio et al., 2017). Similarly, postural sway increases as the complexity of the balance task increases (Mademli et al., 2021). The tandem steps and the FMS lunge were more complex tasks than either standing or walking. The tandem and lunge produced significant increases in sway compared to the single leg stance at baseline. In previous studies restricted arm movements during single leg stance, tandem stance or backward beam walk produced an increase in mediolateral sway (Muehlbauer et al., 2022; Objero et al., 2019). During the tandem steps and single leg stance activities in this study, there were no restrictions on arm movements, but the lunge activity fixated the arms by holding the pole behind the back, as per the FMS protocol. During the lunge there was a correspondingly large mediolateral vector magnitude acceleration compared to the other tests. Therefore, the increasing sway from single leg stance to the tandem steps and further to the lunge in this study was not unexpected.

Mediolateral and anteroposterior sway in near fallers was expected to be significantly more than non-fallers during single leg stance, comparative to older adult fallers compared to non-fallers (Oliveira et al., 2018). However, there was no significant increase in mediolateral sway during single leg stance in near fallers compared to non-fallers. In studies of tandem walk, older adults have a wider base of support and increased sway compared to young adults (Virmani et al., 2018). In similarly narrow support-based investigations of beam walking there was significantly increased mediolateral sway in fallers (Sidaway et al., 2022). However, increased mediolateral sway was not apparent during the tandem steps in the near fallers in this study. During the lunge activity, there was no significant difference between near fallers and non-fallers either.

In previous research, angular velocity sway measures were more useful than acceleration amplitudes to distinguish fallers from non-fallers (Kozinc et al., 2020) but this did not translate to the differences between near fallers and non-fallers for baseline testing. Overall, there was no significant difference in any of the sway variables during any of the baseline tests between the near fallers and non-fallers. This new finding highlights the similarities between the near fallers and non-fallers during baseline tests.

Distraction

The next hypotheses involved distracted conditions. The first hypothesis regarding distraction effects on sway was that distraction would cause increased sway across the cohort compared to the same tests at baseline. This hypothesis was accepted. Distracted balance testing aims to direct attention away from the balance activity. More complex cognitive tasks create increase postural sway (de Barros et al., 2021). This was demonstrated by sway vector magnitude acceleration, as well as root mean square acceleration in mediolateral and vertical directions, significantly increasing for all three balance tests compared to baseline, consistent with the literature (de Barros et al., 2021; Hadad et al., 2020; Tweel et al., 2022).

Distraction, or attention focused on something other than the balance activity, is a known risk factor for falls and near falls (Bitzas et al., 2022). The theory was applied to near fallers, with the hypothesis being that near fallers would sway significantly more than non-fallers during distracted balance testing. A degree of executive control is required for any balance activities (Woollacott & Shumway-Cook, 2002) although practising or familiarity with a cognitive task reduces the dual task cost on the balance activity (Kiss et al., 2018). Conversely, extreme difficulty with the cognitive task could be considered mentally fatiguing, thereby also increasing postural sway (Brahms et al., 2022; Hachard et al., 2020, Qu et al., 2020). Executive processing during a dual task reduces the reactive responses to balance (Solis-Escalante et al., 2019). The expected automatic balance control that normally occurs during distracted balance tasks (Saint-Amant et al., 2020) was confounded further by the conscious deliberation required for the lunge and tandem steps in this study. Further, as this was a study investigating balance, participant volunteer bias may, potentially, have prioritised the balance aspect over the cognitive activity (Plummer & Eskes, 2015).

This study provides the first information about the performance of near fallers during distracted balance testing. There have been few studies investigating distracted balance in midlife adults (Herssens et al., 2018) although a recent investigation found that anticipatory postural control was worse in midlife adults than young adults during distracted balance activities (Bech et al., 2022). The available literature on distracted balance tasks investigates mainly older adults (Brustio et al., 2017; de Barros et al., 2021; Ghai et al., 2017; Kal et al., 2022). Other literature evaluating distracted balance compare fallers with non-fallers (Howcroft et al., 2018; Kozinc et al., 2020), diagnostic groups with healthy controls (e.g., Bishnoi & Hernandez, 2020; Purcell et al., 2020;

Tacchino et al., 2020) or at the other extreme, athletes (e.g., Morelli et al., 2020; Pitt & Chou, 2019; Sarto et al., 2020). This study's findings adds to this existing knowledge.

In previous research of distracted gait, prospectively reported fallers had increased anteroposterior and mediolateral variability than their non-faller counterparts (Howcroft et al., 2018). Therefore, near fallers were hypothesised to have increased anteroposterior and mediolateral sway compared to non-fallers during the similar motion forward of tandem steps. Interestingly, near fallers had significantly less angular velocity results around the vertical axis, indicating less trunk rotation, than non-fallers during the distracted tandem steps.

The tandem steps are a useful, simple measure of dynamic gait function (Robertson & Gregory, 2018). During tandem steps, the front leg controls the anteroposterior sway whereas the mediolateral control depends on weight-bearing symmetry between the two legs (Rougier et al., 2019). Hip abductor and peroneal muscle strength control mediolateral sway and therefore the stability of the stance leg, which influences the placement of the opposite foot during walking. However, this control deteriorates with age (Arvin et al., 2018). During any gait movement, the trunk muscles serve to provide a stable core from which to move the arms and legs (Zemková & Zapletalová, 2022). As the base of support narrows with tandem steps, the need for core strength to control trunk movement increases (Calatayud et al., 2015). The current study identified that near fallers had significantly less trunk rotation during tandem steps compared to non-fallers.

Two conflicting theories exist regarding reduced trunk rotation as a contributor to postural sway. The first theory is that the reduced rotation is due to age-related stiffness in the thoracic and lumbar spines (Cenciarini et al., 2010). This, coupled with age-related muscle loss, translates to a passive stiffness in the trunk, a wider base of support and increased falls risk in older adults. More recent literature confirmed these findings, with older adults demonstrating increased trunk rotation stiffness during normal and fast walking than their younger counterparts (Cury et al., 2020). Accordingly, the reduced trunk rotation in the near fallers could be due to age-related trunk stiffness. However, age-related stiffness is unlikely in this cohort. The median age of the near fallers (64 years) was the same as the non-fallers, demonstrating firstly, no ageing effect between groups, and secondly, the midlife median age (40 - 64 years) not older age (≥ 65 years) of both subgroups.

The second theory is that the reduction in trunk rotation is an active, muscular response to minimise movement of the centre of mass. Trunk control provides a vital role in forward translation such as with tandem steps (Bakshi et al., 2020). The active trunk control response has been demonstrated during walking with a distraction task (Howcroft et al., 2018). In that study, non-faller older adults had significantly less trunk rotation than their faller counterparts (Howcroft et al., 2018). This concept of muscular control of trunk rotation as a stabilising element was also found more recently during walking in young adults (van den Bogaart et al., 2020). There was reduced sway irregularity during standing tasks when core stability had been specifically trained (Szafraniec et al., 2018). Specific core muscle training also improved movement efficiencies in functional and power-related changes in posture (Sasaki et al., 2019). These findings are contradictory to the hypothesis that near fallers would have less postural control and therefore increased sway compared to non-fallers. It appears that the reduced trunk rotation in near fallers may be associated with the higher physical function and general health scores in the SF36. Near fallers may not be the imminent fallers that were expected in this study, but may be a cohort of active, physically high functioning adults whose higher core stability prevented the falls that occurred in other participants. The connections between sensorimotor function and the cognitive functions during distracted balance in midlife and older adults needs further investigation (Brahms et al., 2022), particularly in the near faller population.

Fatigue

Following the rejected hypothesis of increased distracted sway in near fallers compared to non-fallers, the next set of hypotheses related to fatigued sway. The literature states that aerobic exercise, such as the Incremental Shuttle Walk Test (Chae et al., 2022), has a more detrimental effect on postural sway than anaerobic exercise (Güler et al., 2020). Also, previous research identified that reductions in muscle power in older adults negatively affect balance and gait (Byrne et al., 2016; Boyas et al., 2019). This information led to the hypothesis that, across the whole cohort, fatigued balance testing would produce increased sway compared to baseline testing. This hypothesis was partially accepted: all the sway acceleration variables increased significantly during fatigued testing compared to baseline, aligning with the literature (Ghamkhar & Kahlaee, 2019).

However, most of the angular velocity results from fatigued testing were not significantly different to baseline across the cohort. For all participants roll angular velocity, interpreted as hip abduction and trunk side flexion, did not significantly change from baseline during any of the three fatigued balance tests. The lack of significant change from baseline to fatigued activities indicated that the stabilising muscles of the trunk, the weight-bearing stability of the standing leg, and lateral ankle control were not significantly affected by fatigue. The two significant angular velocity increases from baseline due to fatigue were increased trunk flexion during fatigued single leg stance, and increased trunk rotation during the fatigued tandem steps. The significant increase in trunk flexion aligns with the literature on increased anteroposterior sway due to fatigue (Bruniera et al., 2013; Youm et al., 2014). The increased trunk rotation due to fatigued tandem steps suggests the trunk was less able to act as a stabilising central core when the body was fatigued. Fatigue reduces the strength and power of muscle contractions (Enoka & Duchateau, 2016) which could explain the reduced core stability in the whole cohort. Recent research identified bigger changes in trunk and pelvis anticipatory postural adjustments than in leg muscles due to fatigue (Lyu et al., 2021) which may account for this result across the whole cohort.

The second hypothesis relating to fatigued sway was that near fallers would sway more compared to non-fallers during fatigued balance tests. During the lunge and tandem steps there was no significant difference between the near fallers and non-fallers in any direction of acceleration or any plane of angular velocity. During fatigued single leg stance, however, near fallers produced significantly less trunk rotation than non-fallers, suggesting rotational core stability was less affected by fatigue in near fallers. This presents a similar situation to the distracted tandem steps where near fallers had significantly less trunk rotation core stability than the non-fallers. As trunk muscle endurance is an important contributor to single leg stance stability (Ghamkhar & Kahlaee, 2019), it would suggest that the near fallers have better core stability than the non-fallers. Again, these findings contradict the premise that near fallers would present similarly to the fallers depicted in the literature. These new findings of strong core stability are key factors in the near faller cohort, and this will be explored further in the context of the prediction findings.

Prediction

The final hypothesis was that at least one of the acceleration or vector magnitude variables would predict who would become a near faller. The prediction model was built on demographic, biometric, experiential and sway variables. Inertial sensor variables of angular velocity, root mean square acceleration and vector magnitude acceleration were not independent in predicting near falls and did not moderate the model. Therefore, the last hypothesis was rejected. The sway measures taken at discreet time points during balance testing were not useful for predicting the distal outcome of near falls three months later, but neither did they predict the physical function outcome. Hence, postural sway measured during balance tests identified the increased core stability in near fallers but was not helpful for predicting future near falls. While accelerometry sway data collected during near fall simulations in the lab have been shown to be useful in predicting near falls (Pang et al., 2019), they need testing in free-living environments (Fino et al., 2020).

The only predictor of near falls was the SF36 physical function variable, suggesting that near-fallers were successful at preventing a fall in case of a slip or trip. The physical function questionnaire requires self-rated limitations in physical activity, functional strength, community ambulation and activities of daily living (Hays et al., 1993). A higher score, which was the predictor for near falls, was indicative of better functional health and ability. This denotes that the near fallers had few limitations in their daily activities. Physical function is related to successful ageing though physical activity (Bosnes et al., 2019). Ageing-related reduced physical function results from less physical activity (Daskalopoulou et al., 2017) whereas regular physical activity during midlife supports active ageing (Atallah et al., 2018). People who are physically active in midlife were usually physically active as teenagers (Lundell et al., 2019), and active midlife adults tend to continue to be active and have better physical function in later life (Figgins et al., 2021). Particularly for women, regular physical activity in midlife promotes the maintenance of muscle mass and aerobic fitness in later life (Edholm et al., 2021). This study cohort was predominantly female

(74.1%), aged in their midlife (median age 64 years), with high physical function (SF36 physical function median score 89). There was no significant difference between the near fallers and non-fallers for these measures. Near fallers were predicted by high physical function and demonstrated better core stability during distracted and fatigued testing. This highlights the relationships between good physical function, core stability and near falls, which suggests that near fallers are highly physically able, and that the higher ability provides more opportunity for near fall events. The reasons this group sustained near falls but did not fall is possibly due to the higher strength, power and core stability associated with higher physical functioning. Further investigations are necessary to substantiate this hypothesis, by evaluating objective measures of core strength and physical function with near falls.

Physical activity has a direct relationship with muscle power (Muehlbauer et al., 2015), particularly in women after menopause (Straight et al., 2016) as represented by participants in this cohort. The national Physical Activity Guidelines in Australia recommend that adults are active on most days, engaging in 1.25 - 2.5 hours per week of vigorous activity or twice that amount of moderate activity, or an equivalent combination of the two (Department of Health and Aged Care, 2021). Vigorous activity is more likely to build muscle power. Muscle power is more important than strength for predicting function in older adults (Byrne et al., 2016) and is an essential component of postural control (Moura et al., 2020; Stolzenberg et al., 2018). The results of the study indicating higher physical ability and trunk control in near fallers suggests their muscle power, as much as their strength, permitted recovery from their near fall events. That is, when they experienced a near fall, their reactive balance muscle power prevented the near fall from becoming a full fall. Muscle strength, power and reactive balance were not specifically investigated in this study, but future research on these variables in near fallers is essential.

The reasons for near falls provided by the retrospective survey and the prospective diaries varied in detail. Providing participants with a checklist for cause of near fall would ensure consistency for future reporting and coding. However, the cause of a near fall being solely due distraction depends on other observable contributors to the near fall, such as the environment, surface texture or activity being undertaken at the time (Bitzas et al., 2022). Most occurrences of near falls in this study were from trips or stumbles, similar to the older adult near faller investigated by Arnold et al., (2007). In the working population, trips are common when there is limited light availability (Li et al., 2020) and in high-risk work conditions such as building sites or transport hubs (Larue et al., 2021). This differs from the near falls caused by slips, previously reportedly as the most common cause for falls in older adults (Luukinen et al., 2000) and in working populations such as cleaners (Bitzas et al., 2022). The typical recovery from trip is a righting reaction of increased trunk flexion (Handelzalts et al., 2020) and a recovery step (Qu et al., 2020). Recent research indicated that reactive arm movements do not influence recovery from a trip (Gholizadeh et al., 2020), which differs to the observations that led to renewed definitions of near fallers (Maidan et al., 2014).

The definition and rationale for the label of 'near faller' needs to be refined further. For self-reported near falls, using a checklist of the compensation strategies suggested by Maidan may be a start - unplanned movement of arms or legs; unplanned change in step length or speed; trunk tilt forwards; or lowering the body toward the ground (Maidan et al., 2014). A different approach is required for future studies, such as assessing multiple mobility metrics together (Buchman et al., 2020), and gathering sensor data from free-living conditions that are more likely to pick up subtle changes to identify near falls (de Venuto & Mezzina, 2020; Fino et al., 2020; Handelzalts et al., 2020; Kelly et al., 2022). Objective measures would also reduce the burden on future participants to complete regular diary entries.

Mechanisms of postural control, and risk factors for near falls and falls, are complex and diverse, so this study took a multifactorial approach to investigating the near falls. While falls are multifactorial, near falls provide further abstruseness. Each of the risk factors for falls and near falls has low predictive value. The combination of risks were difficult to predict, and potentially influenced by external factors which were not examined in this thesis. Multiple aspects, particularly physical function and core stability were related to near falls. As the evidence in the literature indicates that near fallers become fallers (Nagai et al., 2017), there must be a change in physical function and core stability to become a faller. As fatigue affects power and balance control (Bohrer et al., 2022) there must be a change related to muscle power, likely in reactive balance and core stability, that would identify the transition from a near faller to a faller. As perturbation training improves balance reactions and improves recovery from trips (Okubo et al., 2019; Wang et al., 2020) and slips (Allin et al., 2020), this is recommended as an essential component of balance related physical activity. The proposal of a trajectory from near faller to faller being dependent on physical ability and core stability requires confirmatory studies. Further work is required to predict near fallers.

Limitations

Causality cannot be established by this longitudinal study, as the observations and self-reports of many variables and underpowered sensor data do not permit causality to be determined. The sample was self-selected which introduces selection bias in itself. This thesis reports on the volunteer group who presented for testing and is not generalisable to the whole population. Further, misclassification was possible but was mitigated by the prospective data collection. Misclassification is always a true limitation of subjective data, when reliance is solely on what is reported. The multiple tests were accounted for by incorporating only the prospective data for analysis, using Bonferroni adjusted significant values for multiple tests and selecting only the significant findings for the regression analysis. However, the study does provide new knowledge on midlife and young older adult near fallers, their demographics, balance capabilities and their sway measures.

The sample of sway data was less than the demographic numbers, due to sensor dropout. The sensor chosen for this study was accessible, affordable, rechargeable and the best available at that time. The community venues were selected to facilitate recruitment and make it as convenient as possible for participants. However, sway measures are inherently variable (Kang et al., 2019). Further, the Wi-Fi connections in the community settings, and the Bluetooth connections between sensor and receiver, were also variable. Future research may need to engage a direct cable link to a data logger (e.g., Reynard et al., 2019). The remaining variance may be explained by artefacts in the sample, attributed to the small sample size, self-selected sample, measurement error, self-report bias, and the inexact nature of near falls.

Chapter Summary

This chapter demonstrated new knowledge on a group of midlife and young-older adult near fallers who were identified by their high physical function and distinct from non-fallers by their higher core stability. These findings suggest that near fallers have greater physical ability than their non-fallers counterparts which, in turn, provides more opportunities for slips, trips, and missteps. Recovery from these events is possible due to good core strength and reactive muscle power. The next chapter considers these new findings in relation to clinical implications and future directions for research and policy.

CHAPTER 9 CONCLUSION

This study was the first to explain postural sway in near faller, midlife adults. It contributes new knowledge by explaining the differences between near fallers and non-fallers. These findings will be explored in terms of clinical implications and future directions.

At the end of the diary collection period, the defining difference between the near fallers and fallers was the near fallers' ability to recover balance after a slip, trip or misstep. This suggests that their recovery from a trip or other near fall was due to better reactive balance. The distinguishing difference between the near fallers and non-fallers was their core stability, as indicated by reduced trunk sway during fatigued and distracted balance conditions. The only predictor for near falls was high physical function. This suggests that near fallers were a physically able group who, because of their high physical function, potentially were exposed to more frequent opportunities for slips, trips and missteps than the non-fallers.

From a clinical perspective, it is essential to encourage daily physical activity to meet the Australian guidelines. Thirty minutes of moderate exercise each day attains the target activity. Physical activity can be incidental or purposeful, prescribed as an adjunct to therapy or as a primary target therapy. Health and exercise staff who are engaged with midlife adults in health promotion, wellness provision, work fitness screening as well as therapy settings can incorporate physical activity approaches. These are not difficult to refine for individual circumstances and do not necessarily require great financial outlay to the client. Other than encouraging physical activity, specific exercises to build muscle power will maximise the speed of muscle contraction required to recover balance if a slip, trip or misstep occurs. For example, tip toes, squats and lunges done at different depths and speeds will generate power in the lower limbs for righting reactions. Also, balance and strengthening activities that require both eccentric and concentric muscle control are important to permit the ankle and hip anticipatory postural adjustments. Adequate muscle length is an inherent component of eccentric control. Incorporating stretching, with particular attention to calf length will permit adequate range of movement for balance recovery. Training reactive balance skills provides a complement to the power training. Perturbation training integrates components of hip flexion/extension and step recovery patterns on unstable surfaces and with external perturbations. Lastly, creating and maintaining a stable core, which is essential for balance stability, are beneficial inclusions for training regimes.

This research has opportunity to influence policy. Health, social and ageing policies across all sectors encourage physical activity for healthy living, as well as healthy ageing. Healthy ageing needs to be considered at all ages, not only in older age. Policies ensuring safe access to public facilities encourage regular, incidental and intentional physical activity for all ages. The same principles can be incorporated into health and wellness in the workforce. Core stability screening alongside physical activity measures would indicate who may be at risk of not recovering from a slip or trip incident in the workplace. Preventive programs, as described above, can be implemented for work health and safety reasons, when the relevance of near falls and near misses may be somewhat obscured. The personal approach can complement the environmental risk management policies to provide a more comprehensive risk assessment and mitigation plan.

The cross-sectional nature of data from this study was unable to provide a cause for the near falls, so further research is required. Future research will require a fully powered sample to provide interventions for power training, reactive balance mechanisms and core stability in non-fallers to evaluate if these are the key variables for preventing falls. Near falls were regularly documented, indicating that near falls do occur, and often. However, the nuances of near fall information would be enhanced in future research with a more concrete definition of a near fall, for example using the checkbox of compensatory movements or more objective measures using inertial sensors. Acceleration data from free living environments, rather than a specific testing site, would provide real time, longitudinal data of near falls. This would provide ecological validity in contrast to the current lab-based studies. Free living accelerometry data would also provide physical activity metrics to inform the physical function findings. This approach would provide the next level of information to explore and inform the transition from near falls to falls. The second potential setting for near fall investigation is the workplace. The occupational hazards of slips and trips are well documented, but the intrinsic capacity of the midlife working population in to avoid occupational slips and trips is poorly understood. Providing insight of near miss risk mitigation from both environmental and personnel perspectives would inform work health and safety research, practice and policy.

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APPENDICES

A Manuscript: Scoping Review of Systematic Reviews

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Classification of Balance Assessment Technology: A Scoping Review of Systematic Reviews

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Abstract. Accurate assessment of postural balance is necessary to identify and measure falls risk, inform clinical practice, determine efficacy of treatment, and ultimately falls prevention. The aim of this scoping review was to identify gaps and inform practice, research and policy. There are a multitude of technologies available for assessing balance and no one that meets the requirements of every situation. Force plates had provided the gold standard technology for measuring centre of pressure variables as the cornerstone of balance assessment. Inertial measurements units are now considered as valid and reliable, however inertial sensors in smartphone require further refinement to measure with the same degree of accuracy. Fusion systems combine wearable and non-wearable technology in formal gait labs but also gaming. The flexibility provided choice of wearable, non-wearable and fusion systems meets most clinical and research requirements.

Keywords. Postural balance, technology, reproducibility of results, systematic review

Introduction

Accurate and reliable balance assessments are necessary to identify and measure falls risk, inform clinical practice, determine efficacy of treatment and ultimately prevent falls. Balance is a multi-component, complex construct. It incorporates static balance, the maintenance of body position at rest and dynamic balance, the maintenance of postural stability while the body is in motion, including gait. Internal and external forces alter balance. Some may be anticipated, requiring proactive control; others are unexpected, requiring reactive control [1]. Different tools and methods are available to measure these various aspects of balance.

Balance assessment using instrumented testing devices evaluates static position and dynamic movements [2]. Force platforms measure displacement of centre of pressure (COP) in antero-posterior and medio-lateral directions as a function of time, such as displacement or velocity, or frequency in Hz.[3]. Balance monitoring and assessment for falls prevention has used three categories of technology [4]: 'wearable', 'non-wearable' and 'fusion', the combination of the two. Wearable systems are sensors incorporated into objects, clothing, or footwear, worn on or by the subject. They may consist of inertial sensors combining one or more components of accelerometer, gyroscope, inclinometer, barometer or magnetometer, and may also contain pressure or vital sensors. Fit bits, smart clothing and pressure sensor insoles in shoes are common examples. Non-wearable systems are fixed items that the subject moves on or around. They evaluate kinetic and kinematic activities through pressure receptors, force plates, cameras and other electromagnetic tracking systems. To date, stabilometry, posturography and force plates have provided the gold standard in measuring key components of balance such as centre of pressure excursion and ground reaction forces. Fusion systems are useful to evaluate dynamic balance and gait by using a combination of wearable sensors, non-wearable cameras and/or force plates to measure movement.

Different methods to assess and monitor components and overall balance have become available as advances in technology have occurred. Increasingly, fusion systems are incorporated in virtual reality which use a human–computer connection to assess the user's response during immersion in simulated and artificial environments. The degree of immersion varies from low, when an image is portrayed on a screen, to semi-immersive or augmented where the virtual images overlaying something real, to fully immersive where the user becomes part of the environment. Nintendo Wii and Xbox Kinect are considered fusion systems because the Wii is a force platform with an infrared camera in the hand control [5] and Xbox Kinect consists of a video camera, depth sensor and projector to reconfigure the subject's image on the screen [6]. Virtual reality gaming for assessing dynamic aspects of balance have become increasingly portable, affordable and accessible. Individuals in the community can use gamification and virtual reality to improve physical and cognitive skills that support balance at home.

Clinicians have access to a range of technologies to measure and assist people with declining or challenged balance. While increased choice enhances capability to measure balance, understanding the accuracy and reliability of the technology is essential to ensuring safe, meaningful and clinically relevant outcomes. The principle aim of this review was to summarise the current literature to identify gaps and inform practice, policy and research into the assessment of balance using technology [7]. This scoping review identified systematic reviews using technology to assess balance and has synthesised the evidence for their use based on the reliability and reproducibility of the data collected.

Methods

Design, Search Methods, Eligibility and Data Extraction.

The methodology followed scoping review guidelines [7], with no quality analysis undertaken as the intent was to scope the literature and clarify ideas by mapping the available evidence [8]. The search was limited to the past five years due to the rapid expansion of technology. The PRISMA extension for scoping reviews was used as a reference [9]. The search strategy was refined with the support of an expert librarian using MeSH terms and keywords relating to 'postural balance', 'technology', 'reproducibility of results' and 'systematic review' (see Appendix 1). Five databases were comprehensively searched (Medline, Embase, CINAHL, SCOPus and PubMed) from 2013 to 26 August 2019 by two independent reviewers (see Figure 1, PRISMA flowchart [10]).

Inclusion criteria were systematic reviews examining technology to evaluate postural balance in any age or diagnostic group, written in English. Exclusion criteria were balance other than postural (e.g., work-life balance); stability other than postural stability; equilibrium not related to posture; gait only; balance not measured by technology; not systematic review; fall detection methods. A Microsoft Excel spreadsheet was used to assemble data extracted from the selected full text articles.

Two reviewers independently extracted data from 10 articles to compare and agree on detail for consistency before the remaining data extraction was completed. Criteria for data extraction included author, year of publication, title, number of studies included, databases searched, MeSH headings or keywords, date range, inclusion and exclusion criteria, aims or purpose, methodology, population of interest, intervention, comparison group, outcomes, balance assessment tool, technology utilised to assess balance, bias, strength of evidence, limitations, key findings, clinical relevance and gaps identified.

Findings

Search and Selection Strategy

The search identified 792 articles and 360 duplicates were removed. Of the remaining 432 articles, 44 met the inclusion criteria (see Figure 1, Screening process).

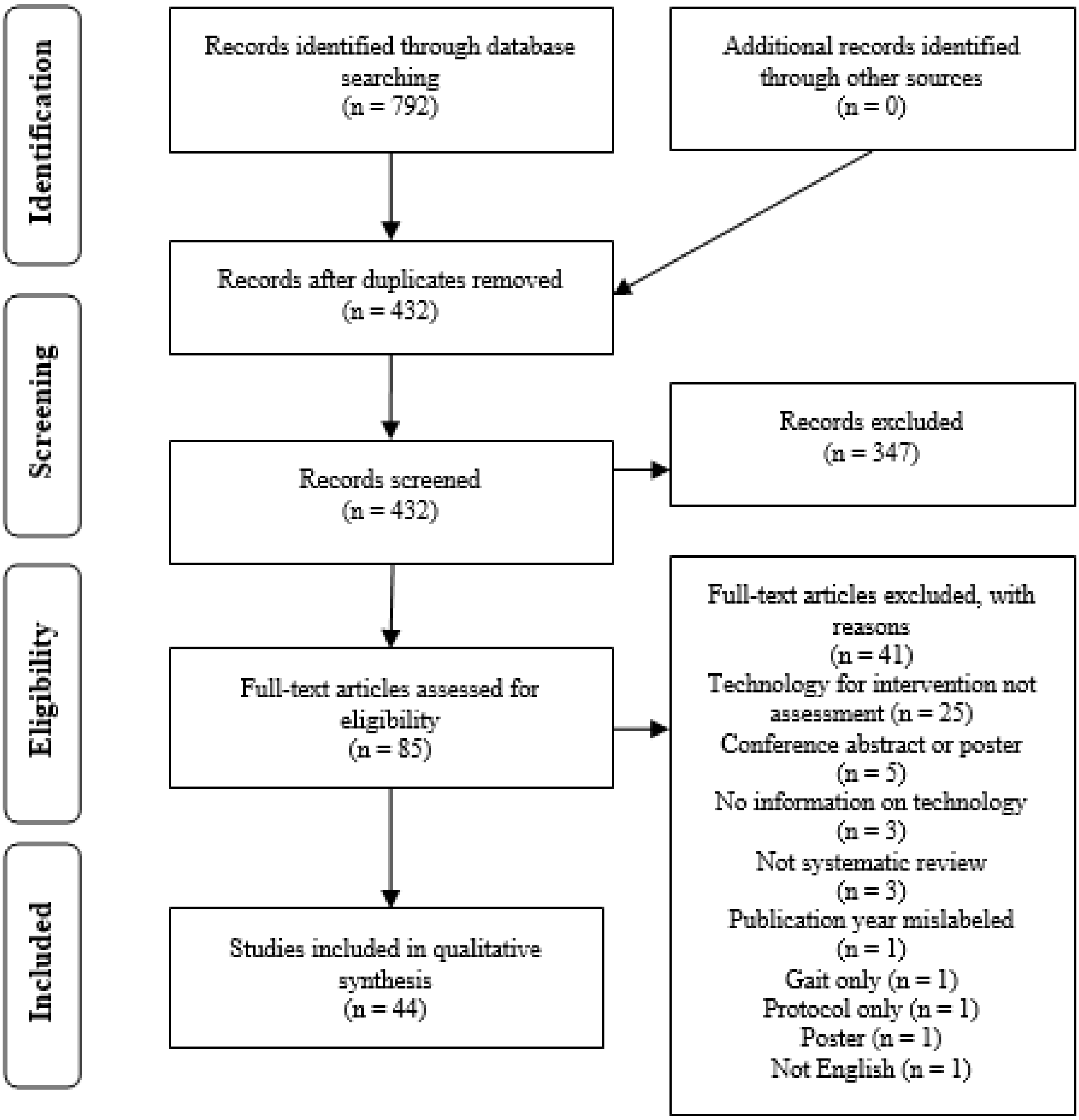

Figure 1. Screening process

Included Reviews and Characteristics

An overview of review findings is provided in Table 1, Summary of Included Studies. All 44 reviews were conducted over the past 4 years. The number of databases searched in the reviews ranged from two [11, 12, 13, 14, 15] to seven [16, 17, 18, 19] with the most commonly searched being PubMed/Medline (97.7%), EbscoHost including CINAHL (45.5%), Embase (40.9%) and Web of Science (40.9%). The IEEE database was searched in only one study [20]. The number of studies in each review varied from 5 [15,21] to 115 [22] with a mean of 19.8. Strength of evidence was based on triangulation with co-authors; scoring papers within the review against criteria such as PEDro [23] or Downs and Black [24] quality checklists; evaluating heterogeneity, effect size e.g., Hedge’s g or Cohen’s d, weighted or standard mean difference, or calculating confidence intervals; and performing meta-analysis. Multiple populations, interventions, assessment methodologies and clinical measures were identified; see Table 1 Summary of included studies.

Table 1 Summary of included studies

Author, Year Number of studies [Reference]	Population (P) Intervention (I) Comparator (C)	Outcome (O) Clinical measure (CM)	Key findings
Bahureska et al., 2016 N=14 [25]	Mild CI (P) Dual tasks (I) No CI (C)	Gait, COP (O) COP, EO, EC (CM)	Single task slowed gait velocity, dual task arithmetic highest sensitivity
Büttner et al., 2019 N=26 [16]	Sporting ABI (P) Walking, static balance, dual tasks (I) No injury (C)	Balance, gait (O) CoM, COP (CM)	Dual task slowed gait, increased frontal CoM sway
Casuso-Holgado et al., 2018 N=11 [46]	MS (P) Balance, gait training, VR (I) Usual care (C)	Balance, gait, function (O) COP, BBS, POMA, Gait, ABC, FRT, FES, SLB, CES, FSST (CM)	VR more effective than no treatment; not sig compared to usual care; not conclusive for gait
Cheok et al., 2015 N=6 [17]	Adult stroke (P) Wii (I) Usual care (C)	Balance, gait, function (O) FIM, BI, TUG, BBS, FRT, FES (CM)	Wii more effective than usual care for TUG but not BBS or FIM; not sig compared to usual care; good retention
Chen et al., 2016 N=9 [56]	Stroke (P) VR (I) Usual care (C)	Balance, gait, function (O) TUG, BBS, 2MWT, FAC (CM)	VR moderate effect with usual care in chronic stroke; less so with acute or subacute stroke
Cieslik et al., 2019 N=15 [26]	CI (P) Static balance, force plate (I) Young (C)	COP variables (O) BBS, TUG, BESTest, COP (CM)	Need to clarify level of CI and posterior stability
Clark et al., 2018 N=25 [47]	Adults, children (P) Wii (I) Force plate (C)	COP variables (O) FT, SLS, EO, EC, foam (CM)	Wii concurrent validity with commercial force plates; reliable for static standing computerised posturography; low cost, portable, open access
Corbetta et al., 2015 N=15 [48]	Adult stroke (P) VR (I) Usual care (C)	Balance, gait, function, adverse events (O) 6MWT, 10mWT, BBS, FRT, POMA, TUGT, COP (CM)	VR rehabilitation improved walking speed, balance and mobility; VR improved balance > usual care; relatively safe

de Amorim et al., 2018 N=10 [49]	Older adults (P) VR (I) Usual care (C)	Balance, gait, function (O) BBS, TUG, SLS, POMA, ABC, COP, gait (CM)	Improvements in balance, mobility, flexibility, gait and fall prevention
Dewar et al., 2014 N=45 [27]	CP, brain injury, children (P) TT, VR, reactive balance, visual biofeedback (I) NDT (C)	Balance, function (O) FRT, TUG, BBS, GMFM st/sitt, COP, PBS (CM)	Hippotherapy, treadmill, trunk-targeted, reactive balance, gross motor task training moderate; NDT, VR, visual biofeedback, FES low; not resisted or UL exercises
Dominguez Ferraz et al., 2017 N=12 [50]	PD (P) Wii (I) Healthy (C)	Balance, gait, function (O) BBS, TUG, DGI, COP Rhomberg, POMA, BESTest (CM)	Wii intervention may improve balance and mobility in adults with PD
Dos Santos et al., 2015 N=5 [21]	Adult stroke (P) Wii (I) Usual care (C)	Balance, function (O) BBS, TUG, COP (CM)	Little evidence of effectiveness of Wii treatment in patients with sequelae caused by a stroke
Dufvenberg et al., 2018 N=18 [28]	AIS (P) Postural stability (I) Healthy (C)	Cobb angle, COP variables (O) COP (CM)	Moderate quality evidence for decreased postural stability in AIS measured as COP parameters
Dumont et al., 2015 N=14 [29]	Adult stroke (P) Not reported (I, C)	Balance (O) BBS, TUGT, 6MWT, 3mWT, POMA, DGI, 10mWT, gait, COP (CM)	No assessment tool allows the evaluation of both static and dynamic balance in stroke survivors
Gebel et al., 2018 N=17 [11]	Young (P) Stability (I) Usual care (C)	Balance (O) Flamingo, SEBT, Standing Stork Test, COP (CM)	BT improves balance irrespective of age, sex, training status, setting and testing method
Gobbo et al., 2014 N=8 [37]	Older adults (P) Dual task (I) Not reported (C)	COP variables (O) FRT, ABC, 6MWT (CM)	Exercises provided limited benefit in static or dynamic balance during dual tasks
Gordt et al., 2017 N=8 [43]	Adults (P) BT (I) Usual care (C)	Balance, gait, function (O) TUG, BBS (CM)	Wearable sensor training improved static and dynamic balance
Hubble et al., 2015 N=26 [38]	PD (P) Sensors (I) No sensors (C)	COP variables (O) Quiet stance, TUG, gait, tandem, Rhomberg, EO, EC (CM)	Wearable sensors detect standing balance and walking stability differences in PD and controls
Johnston et al., 2019 N=47 [39]	Sporting, young adults (P) IMU (I)	Validity, reliability (O)	Wearable inertial sensors provide valid, reliable measures of postural control

	Not reported (C)	BESS, SOT, YBT, DPSI, CTSIB, SEBT, (CM)	
Juras et al., 2019 N=20 [51]	Stroke, PD, CP (P) VR (I) Usual care (C)	Balance, gait, function (O) FRT, TUGT, BBS, POMA, PDRS, 10mWT, STS, CBM (CM)	Training in a virtual environment showed significantly better results.
Kamieniarz et al., 2018 N=32 [12]	PD (P) Motion analysis, posturography (I) Not reported (C)	UPDRS (O) POMA, ABC, COP (CM)	Results are contradictory; Variability of scales and tests to assess balance in people with PD
Kümmel et al., 2016 N=6 [30]	Knee OA (P) BT (I) Healthy (C)	Balance (O) Time at stability, COP (CM)	Task-specific improvement after balance training; limited or no effect on non-trained balance tasks
Lawson et al., 2015 N=21 [31]	Knee OA (P) Balance (I) No knee OA (C)	COP variables (O) COP (CM)	People with knee osteoarthritis exhibit altered postural control
Lesinski et al., 2015 N=23 [13]	Older adults, healthy (P) BT (I) Other exercise (C)	COP variables (O) COP (CM)	BT improved proxies of steady-state, proactive, and reactive balance; effective BT modalities needed to improve balance in healthy older adults
Li et al., 2016 N=16 [32]	Stroke adult (P) VR (I) Usual care (C)	Balance, gait, function (O) BBS, BBA, POMA, TUG, FRT, ABC, COP (CM)	Supports the use of VR to improve balance after stroke
Low et al., 2017 N=23 [18]	Older adults (P) Activity (I) Usual care (C)	COP variables (O) COP (CM)	Balance exercise interventions improve postural control compared to strength or other exercise
Luque-Moreno et al., 2015 N=11 [33]	Stroke, lower limb rehab (P) VR (I) Usual care (C)	Balance, gait, functional performance (O) 6MWT, FAC, MMAS, fMRI, ABC BPM, STS, SIS; 10mWT (CM)	VR provided significant improvement on gait speed, balance and motor function
Ma et al., 2016 N=17 [44]	Adults (P) Wearable sensor (I) No system (C)	Balance, gait, COP variables (O) Rhomberg, tandem, LOS, SEBT, gait, BBS, TUG(CM)	Most wearable sensors are effective in assessing static and dynamic balance
Manlapaz et al., 2017 N=16 [19]	Older adults (P) Wii (I) Usual care (C)	Balance, gait, Wii scores (O) BBS, TUG, MDRT, FABS, FRT, FES, POMA, STS, SLS, COP (CM)	Outcome measures inconsistent across studies; Wii may be used as an outcome measure using Wii Bubble and Wii Score features

Pang et al., 2019 N=9 [45]	Older adults (P) Not reported (I, C)	Falls, near falls (O) COP (CM)	Wearable devices have high accuracy to detect laboratory induced near falls
Petro et al., 2017 N=63 [34]	Not described (P) Dynamic balance using IMUs (I) Static balance (C)	Balance (O) COP (CM)	Dynamic balance assessment methods complement static assessments for studies involving postural control.
Pinho et al., 2019 N=9 [41]	Healthy adults (P) Balance apps (I) Usual care (C)	Balance (O) BESS, SLS, Romberg, tandem Romberg, SOT, COP (CM)	Mobile devices may have accuracy to assess postural balance, poor quality studies, not strong evidence.
Puh et al., 2019 N=21 [20]	Adults (P) Not reported (I, C)	Balance, function (O) POMA, ABC, COP (CM)	Support for Kinect cameras for limited temporal, spatial, and kinematic measures.
Rodrigues et al., 2014 N=16 [55]	Older adults (P) Exergaming (I) Not reported (C)	Balance, gait, function (O) TUG, POMA, ABC, grip, SPPB, COP, FTSTS, 6MWT, SRT, gait (CM)	VR gaming did not improve the musculoskeletal function of elderly adults
Roeing et al., 2017 N=13 [42]	Healthy adults (P) Static balance (I) Usual care (C)	COP variables (O) TUG, STS, BBS (CM)	Smartphone balance apps still in development and initial testing phases
Ruff et al., 2015 N=11 [14]	Orthopaedics (P) Exergaming (I) Usual care (C)	Balance, function, COP variables (O) TUG, COP (CM)	Exergaming provides a low-cost, accurate and reliable clinical assessment tool in orthopaedics
Sun et al., 2018 N=33 [40]	MS (P) Sensors (I) No sensor (C)	Balance, gait, function (O) T25FW, 6MWT, TUG, push/release (CM)	Sensor-based assessments were highly accurate in comparison with reference measures
Tahmosybayat et al., 2017 N=11 [52]	Healthy older adults (P) Exergaming (I) Usual care (C)	Balance, function, COP variables (O) FRT, FES, ABC, POMA (CM)	Exergaming less effective than alternative BT; exergaming may permit realistic assessment of ADL
Tahmosybayat et al., 2018 N=18 [53]	Healthy older adults (P) Balance (I) Not reported (C)	Balance, function (O) Not reported (CM)	Exergaming aligned 5 of 9 SFPC; Equivalent response to external perturbation; mimic changes in sensory context
Tally et al., 2017 N=8 [35]	Stroke (P) TT (I) Usual care (C)	Balance, function, COP variables (O) BBS, LOS, COP (CM)	TT is a beneficial intervention for balance dysfunction in chronic stroke.

Tariq, 2016 N=5 [15]	MS (P) Wii (I) Usual care (C)	Balance, gait, function (O) BBS, TUG, FSST, COP (CM)	Wii balance games improve static more than dynamic balance in mild to moderate level of disability
Tripette et al., 2017 N=115 [22]	Not described (P) Wii (I) Usual care (C)	Balance, gait, function (O) BBS, TUG, ABC, COP (CM)	Wii fit good for prevention of metabolic disorders; provides similar response as conventional therapy
Veis-Karami et al., 2019 N=10 [36]	AIS (P) Spinal bracing (I) Not reported (C)	COP variables (O) COP, LOS, RWS, WBS (CM)	AIS have increased COP excursion
Vogt et al., 2019 N=16 [54]	LL injury (P) VR (I) Force plate (C)	Balance (O) SEBT, SLS (CM)	VR balance interventions equally effective compared to traditional balance training

Key: 2MWT, Two-Minute Walk Test; 6MWT, Six-Minute Walk Test; 10mWT 10-Meter Walk Test; ABC, Activities-specific Balance and Confidence scale; ABI, Acquired Brain Injury; ADL, Activities of Daily Living; AIS, Adolescent Idiopathic Scoliosis; BBA, Brunel Balance Assessment; BBS, Berg Balance Scale; BESTest, Balance Evaluation Systems Test; BI, Barthel Index; BPM, Balance Performance Monitor; BT, Balance Training; CBM, Community Balance and Mobility scale; CI, cognitive impairment; COP, Centre of Pressure; CP Cerebral Palsy; CTSIB, Clinical Test of Sensory Interaction and Balance; DGI, Dynamic Gait Index; EC, eyes closed; EO, eyes open; FABS, Fullerton Advanced Balance Scale; FES, Falls Efficacy Scale; FIM, Functional Independence Measure; FAC, Functional Ambulatory Scale; FRT, Functional Reach Test; FSST, Four Square Step Test; FT, feet together; IMU, inertial measurement unit; LL, lower limb; LOS, Limits of Stability; MDRT, multidirectional reach test; MMAS, Modified Motor Assessment Scale; MS, Multiple Sclerosis; NDT, neurodevelopmental training; OA, Osteoarthritis; PBS, Paediatric Balance Scale; PD, Parkinson's Disease; PDRS, Parkinson's Disease Rating Scale; POMA, Performance-Oriented Mobility Assessment; SFPC, Systems Framework for Postural Control; SLS, single leg stance; SOT, sensory organization test; SPPB, Short Physical Performance Battery; SRT, sit and reach test; STS, Sit-to-Stand test; TT, Treadmill Training; TUG, Timed Up Go; UPDRS, Unified PDRS; VR, virtual reality; Wii: Nintendo Wii; WS wearable sensors.

Technologies

Various technologies were used to assess balance. Several reviews used one or more clinical assessment proxies in addition to measures of centre of pressure (COP) sway, displacement, velocity. Results are grouped by classification into non-wearable, wearable and fusion in Table 2 which describes the relative advantages and disadvantages.

Table 2. Advantages and disadvantages with balance assessment technologies.

Classification	Examples	Advantages	Disadvantages
Non-wearable systems [12, 13, 18, 25, 26, 27, 28, 29, 30, 31, 32, 33, 34, 35, 36, 37]	Lab-based force plates e.g. Biodex, stabilometry Lab-based camera systems, posturography, gait labs	Gold standard, accurate, sensitive, detailed. Measure quiet stance (stabilometry), postural sway measuring COP (posturography)	Expertise required to operate and interpret findings; high cost; limited by space and requiring dedicated lab; variability of computation options, frequencies, observation periods, position protocols, environmental conditions
Wearable systems [11, 38, 39, 40, 41, 42, 43, 44, 45]	Inertial Sensors, Pressure insoles	Inexpensive, multiple sites on the body, unobtrusive, potential to measure many parameters from one IMU	Sampling frequency, body position variability, interpretation
Fusion systems [14, 15, 16, 17, 18, 19, 20, 21, 22, 27, 32, 33, 35, 40, 46, 47, 48, 49, 50, 51, 52, 53, 54, 55, 56]	Game-based force plates with camera technology e.g. Nintendo Wii, Xbox Kinect, IREX	Functional, fun, interactive, engaging and semi-immersive for intervention; concurrent validity compared with lab-based force plate; low cost; portability; open access as development platform feedback to user; targeted sessions	No standard dosage, frequency, interaction time; no studies using headwear Limited protocols, sampling

Several stabilometry and computerized dynamic posturography systems were used in the formal lab-based classification [12, 13, 18, 25, 26, 27, 28, 29, 30, 31, 32, 33, 34, 35, 36, 37]. In these reviews, one or more of medio-lateral and antero-posterior position and range, COP excursion or velocity were measured. In the wearable classification, 3D inertial measurement units (IMUs) were most frequently used with one or more components of accelerometer and gyroscope [11, 38, 39, 40]. Two papers used IMUs integrated in mobile phones [41, 42] and three papers correlated information with a pressure sensor in the shoe [43, 44, 45] with potential to improve static (IMU) or dynamic (plantar sensor) balance in different populations [44].

Fusion systems combined attributes of the wearable and non-wearable componentry and were used in assessing gait, monitoring movement and providing feedback such as gaming. The most common gaming to be identified in the studies was Nintendo Wii [14, 15, 16, 17, 18, 19, 21, 22, 27, 32, 33, 40, 46, 47, 48, 49, 50, 51, 52, 53, 54, 56]. Of the 23 reviews that investigated Wii, only seven used the Wii to assess balance outcomes [15, 16, 40, 47, 48, 51, 53]; the remainder relied on Biodex or equivalent lab-based force plate measurements as independent outcome measures [14, 17, 18, 19, 21, 22, 27, 32, 33, 40, 46, 49, 50, 52, 54] or clinical tools to corroborate [14, 15, 19, 21, 22, 27, 32, 33, 40, 46, 48, 49, 50, 51, 52, 54]. Xbox Kinect was the second most popular option for virtual reality [14, 20, 40, 53, 54, 55, 56] with studies similarly using lab-based assessment [14, 40, 54, 55] often with clinical tools [14, 20, 40, 54, 55] to evaluate balance outcomes. Balance components within the games were quantified and compared to lab or clinical-based outcomes for different age groups. Wii was also used as an outcome measure by using Wii Bubble or Wii Score features [19].

Populations

Healthy participants of different ages were assessed in some studies as the primary population group [11, 13, 18, 19, 20, 22, 30, 34, 37, 39, 41, 42, 43, 44, 45, 47, 49, 52, 53, 54, 55] and other studies as the comparator group [16, 19, 25, 26, 28, 31, 50]. All ambulant ages were included in the systematic reviews; children [27, 51], adolescents [11, 30], adults [16, 20, 30, 39, 41, 42, 43, 44, 49, 54] and older adults [13, 18, 19, 20, 25, 26, 37, 43, 45, 49, 52, 53, 55]. Table 3 shows the healthy population groups assessed by the classification of technology.

Table 3. Balance assessment technologies used in healthy populations

Healthy population	Non-wearable	Wearable	Fusion
Children	FP [11]		Wii [47]
Youth	FP [11, 30, 34]	IMU [39]	
Adults	FP [22, 30, 34, 54]	MP [42] IMU [39, 41, 43, 44, 45]	Wii [20, 22, 47] VR [54]
Older adults	FP [13, 18, 19, 34, 37]	IMU [45]	VR [49, 52, 53] Kinect [55]

Key: FP force platform; IMU Inertial Measurement Unit; MP, Mobile Phone; VR, Virtual Reality

Force platform technology was used to assess all healthy ambulant age groups, however fusion systems and mobile technology were not used to assess youth or children respectively. Wearable sensors were highly accurate and detected differences in static balance and gait stability in younger [44] and older adults [45]. Wii was identified to have concurrent validity with commercial force platforms [47]. Whilst some studies investigated validity [14, 20, 28, 39, 40, 42, 43, 47] and reliability [14, 20, 39, 40, 42, 47] of measurements in healthy populations. It was not in the capacity of this scoping review to evaluate either in detail. Further to assessing balance in healthy populations, balance within specific conditions was investigated (Table 4 Balance assessment technologies used in specific conditions). Fusion systems, usually incorporating a gaming or virtual reality environment, were the most common approach for assessing balance in specific populations, covering all ages from children with Cerebral Palsy (CP) [51] to the frail elderly [20]. Force platform technology was also consistent across the different patient groups; however wearable technology was less common. One study investigated wearable technology to measure balance in stroke, Parkinson's Disease (PD) and the frail elderly [43] and another focused solely on PD [38].

Table 4. Balance assessment technologies used in specific conditions

Condition	Non-wearable	Wearable	Fusion
Musculoskeletal (MSK)	AIS [28, 36] Knee OA [31] LBP [20]		Athletes [16] Orthopaedic [14]
Cognitive Impairment (CI)	CI [25, 26]		Concussion [16]
Neurological conditions	CP [27] Stroke [20, 29, 32, 33] PD [12, 20]	Stroke [43] PD [38, 43]	CP [51] MS [15, 20, 40, 46] Stroke [17, 21, 35, 48, 51, 56] PD [20, 50, 51]
Frailty		Frail older [43]	Frail older [20]

Key: AIS, Adolescent Idiopathic Scoliosis; CP, Cerebral Palsy; LBP, Low Back Pain; MS, Multiple Sclerosis; PD, Parkinson's Disease

Results from the studies indicated that static balance was reduced in concussed athletes [16], in knee osteoarthritis [31] and Adolescent Idiopathic Scoliosis (AIS) [28]. Balance was improved after balance training interventions but not by strength or multicomponent exercises [11, 13, 18, 30]. This extended to older adults improving balance outcomes after using wearable sensors to train static, dynamic and proactive balance. [43]. In addition, children with CP improved balance outcomes when trained with reactive balance activities [27]. Functional dynamic balance was assessed by gait speed which was slowed during single [25] and dual tasks in mild cognitive impairment [26] and concussion [16]. Dual task arithmetic provided the highest sensitivity in the cognitively impaired group [25].

Virtual reality (VR) was more effective than no treatment but not compared to conventional treatment in Multiple Sclerosis (MS) [46]. Whilst one study found little evidence to support VR training with stroke [21], this was countered by the majority of systematic reviews that identified VR as moderately effective in improving walking speed, balance and mobility when added to usual care [17, 32, 33, 48, 51, 56]. Results were also contradictory for balance using VR with older adults. Two of the included reviews found improvements in static and dynamic balance, particularly whole-body sway, stepping and leaning activities, following use of VR [49, 53]. In contrast, two other studies found little benefit when comparing VR to conventional intervention [52, 55] with this population. Training in VR improved balance outcomes for PD and CP [51] but not the musculoskeletal function of elderly adults [55]. Wii was more effective than usual care for static and dynamic balance and functional gait in MS [15, 17] but was inconclusive for functional balance in PD [50] and had mixed response in Stroke [21]. Wearable sensors were accurate and detected differences in static balance in MS [40] and between people with Parkinson's Disease and controls [38]. Wearable sensors were more sensitive than clinical balance assessment tools for sports medicine [39] although sensors within mobile devices required further investigation to provide strength of evidence [41, 42].

Discussion

This scoping review identified and mapped available systematic review evidence about technology-based balance assessment tools. Foci within the systematic reviews varied regarding population groups, type of technology and intervention protocols. While there were overlaps between reviews there were also gaps in the evidence about the use of technology to assess balance and multiple methodologies were presented. Many of the included studies called for increased rigour in research methodology, including randomization, blinding, intervention protocols for frequency, number of sessions, sampling time and assessment. All ages were assessed by the varying technologies; however, acceptability of different technologies has been identified to differ by age [57] and specific population group such as MS [58], stroke [59] and PD [60].

Force plates, stabilometry, posturography and gait labs provided the gold-standard in measuring balance with accurate and comprehensive assessments [61]. However, they were expensive, required dedicated space and expert interpretation of data and the ecological validity of data collected in these environments is questionable. This made them less useful in the clinical setting and more relevant for research investigations [62]. Also, testing protocols were not consistent, using different foot and body positions, sampling times and frequencies, and number of repetitions for each test, which creates challenges with data meta-analysis and generalizing results. All ages, and specific conditions of stroke, cognitive impairment, Parkinson's Disease and musculoskeletal injuries had balance assessed by the fixed assessment categories, although there were no investigations for participants with MS in this category in the past 5 years.

Wearable sensors were increasingly accurate and able to detect subtle differences in static and dynamic balance [63], although there was no consensus on sensor positioning or sampling frequency [40]. Sensors within clothing were not identified in this scoping review although were emerging as a new methodology to assess balance [64]. This emerging field of wearable sensors assessed balance in all ages, but only two studies investigated wearable sensor use in distinct diagnostic groups: Parkinson's Disease [38, 43] and Stroke [43]. The efficacy of wearable sensor use in mobile phone technology was not proven and warranted further exploration [41], a finding consistent elsewhere [65]. Additionally, for mobile phone assessment to be clinically useful, it required the interpretation of the assessment findings by the end user. Further research into the reliability and validity of mobile phones as wearable sensors is warranted, as well as the efficacy of end user interpretation of results.

Fusion systems combined the fixed and wearable componentry. The established gait labs, whilst providing gold standard gait and dynamic activity outcomes, were equivalent to the non-wearable technology for high cost and low convenience and access [66]. Newer fusion systems consisting of gaming provided increasingly portable and flexible options. However, few studies used the gaming fusion system itself as an assessment outcome measure, even though add-on programs were available for measuring sway and COP variables [67]. The reliance on additional lab-based assessment for accuracy or clinical proxy assessments for functional outcomes reduced the efficacy of this gaming sub-group. Virtual reality interventions were considered effective for stroke, PD and CP but inconclusive for older adults with and without musculoskeletal conditions. Other gaming technologies such as oculus rift were not explored at all and yet recent studies identify this as a valid measures of COP sway without requiring additional force plate cross referencing [68]. Similarly, 23 studies did not meet inclusion criteria for this scoping review because they used balance technology solely for the intervention and not assessment. Studies relied heavily on known clinical tools for outcome measures which may have indicated perceived limits in efficacy of the technology. Further research into the reliability and validity of gaming systems to assess COP variables was therefore warranted.

The results of the included studies provided recommendations for clinical practice, particularly that only balance training exercise, rather than alternate modes of exercise, improved balance. Some population groups such as young people with AIS, concussed athletes and adults with OA knees have known balance deficits that warrant early, targeted balance training intervention. Dual task assessments could provide early diagnosis of mild cognitive impairment before covert balance and functional deficits deteriorate to impact daily activities or increase the risk of falls. cha

Limitations

The search for this scoping review was conducted only in the previous five years due to the rapid advances in technology. However, previous systematic reviews investigating technology for balance assessment may have been missed. Further, only papers written in English were included. Reviews which included analysis of gait without specific reporting of balance were excluded however could be a proxy measure of functional dynamic balance. The clinical utility, reliability and validity of the different technologies was not explored in detail.

Conclusion

There are a multitude of technologies available for assessing balance and not one that meets the requirements of every situation. No reviews identified wearable sensors in clothing, nor fusion systems with headpieces. The flexibility provided by fixed, wearable and fusion systems provides advantages and disadvantages that warrant evaluation for specific assessment of physiological and functional contributors to the multi-faceted construct of balance.

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Appendix: Search strategy for Ovid Medline

#	Searches	Results
1	*postural balance/ or *posture/ or *standing position/	38812
2	(balance* or postur* or sway* or stability or equilibrium).ti,kf.	169358
3	("center of pressure" or "centre of pressure").ti,ab,kf.	4348
4	or/1-3 [Balance concept]	188095
5	*technology/ or *biomedical technology/ or *biotechnology/ or *man-machine systems/	28581
6	mobile applications/ or user-computer interface/ or video games/	44337
7	computers, handheld/ or smartphone/	6511
8	wearable electronic devices/ or fitness trackers/	1582
9	Accelerometry/ or Magnetometry/	4703
10	automation/ or robotics/ or artificial intelligence/	57584
11	Photogrammetry/	2387
12	Internet/	69579
13	(technolog* or gerontechnol* or gerontotechnol* or digital* or computer*).ti,ab,kf.	815483
14	(posturograph* or stabilograph* or force plate* or forceplate* or forcedeck* or zebris or force platform or platform* sensor* or pressure sensor* or pressure mat*).ti,ab,kf.	11747
15	(inertial adj2 (sensor* or monitor* or unit* or system*)).ti,ab,kf.	2733
16	(wearable adj2 (device* or wireless or sensor*)).ti,ab,kf.	4661
17	(fitbit or garmin or jawbone or moov or pebble).ti,ab,kf.	1399
18	((body or motion) adj2 sensor*1).ti,ab,kf.	2113
19	(smart phone* or smartphones* or android* or iphone* or ipad* or app or apps or mobile application*).ti,ab,kf.	32482
20	(video gam* or gam* system* or wii or kinect or nintendo or playstation or xbox or balance board*).ti,ab,kf.	5323
21	(oculus or samsung gear or VR).ti,ab,kf.	7830
22	((virtual or augment* or mixed) adj1 realit*).ti,ab,kf.	10390
23	((virtual or augment*) adj2 (environment* or object* or world* or system* or program* or rehabilitation*)).ti,ab,kf.	8207
24	(acceleromet* or gyroscop* or magnetomet* or goniomet* or inclinomet* or baromet*).ti,ab,kf.	26346
25	(robot* or artificial intelligen*).ti,ab,kf.	44783
26	(internet or online).ti,ab,kf.	139509
27	(videocaptur* or video captur* or motion captur* or mo-cap or mocap or vicon or motion sensor* or motion track*).ti,ab,kf.	6817
28	(Photogrammet* or Stereophotogrammetr*).ti,ab,kf.	2602
29	or/5-28 [Technologies concept]	1149962
30	"reproducibility of results"/ or "sensitivity and specificity"/ or "predictive value of tests"/	753942
31	(accuracy or assessment* or measur* or evaluat* or reliabil* or reproduc* or consistenc* or repeatab* or validit* or sensitivity or responsiveness* or clinimetric).ti,ab,kf.	7316939
32	or/30-31 [Tests or measures]	7534493
33	and/4,29,32	9486
34	limit 33 to (english language and yr="2013-Current")	5044
35	exp animals/ not humans/	4611840
36	34 not 35	4915

37	exp Review Literature as Topic/ or exp Review/ or Meta-Analysis as Topic/ or Meta-Analysis/ or "systematic review"/	2629845
38	((systematic or state-of-the-art or scoping or literature or umbrella) adj (review* or overview* or assessment*)).tw.	233836
39	("review* of reviews" or meta-analy* or metaanaly* or "research evidence" or metasynthe* or meta-synthe*).tw.	160866
40	((systematic or evidence) adj1 assess*).tw.	4019
41	or/37-40	2723022

B Manuscript: Systematic Review

sensors

Systematic Review

Inertial Sensor Reliability and Validity for Static and Dynamic Balance in Healthy Adults: A Systematic Review

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Abstract: Compared to laboratory equipment inertial sensors are inexpensive and portable, permitting the measurement of postural sway and balance to be conducted in any setting. This systematic review investigated the inter-sensor and test-retest reliability, and concurrent and discriminant validity to measure static and dynamic balance in healthy adults. Medline, PubMed, Embase, Scopus, CINAHL, and Web of Science were searched to January 2021. Nineteen studies met the inclusion criteria. Meta-analysis was possible for reliability studies only and it was found that inertial sensors are reliable to measure static standing eyes open. A synthesis of the included studies shows moderate to good reliability for dynamic balance. Concurrent validity is moderate for both static and dynamic balance. Sensors discriminate old from young adults by amplitude of mediolateral sway, gait velocity, step length, and turn speed. Fallers are discriminated from non-fallers by sensor measures during walking, stepping, and sit to stand. The accuracy of discrimination is unable to be determined conclusively. Using inertial sensors to measure postural sway in healthy adults provides real-time data collected in the natural environment and enables discrimination between fallers and non-fallers. The ability of inertial sensors to identify differences in postural sway components related to altered performance in clinical tests can inform targeted interventions for the prevention of falls and near falls.

Keywords: inertial measurement unit; postural balance

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1. Introduction

Postural control of balance is essential for keeping upright, moving effectively, and reacting to environmental challenges [1]. Good balance improves quality of life and wellbeing. Conversely, balance deficits can lead to a near fall or fall that may result in physical, psychological, or social consequences and, in some cases, death [2]. Near falls occur due to a loss of balance from a slip, trip or stumble where a fall is avoided "because a corrective action is taken to recover balance" [3] (p. 49). Although near falls are a predictor for falls [4], there is limited research concerning near falls, resulting in an unknown trajectory of the decline from near falls to falls [5]. People living in the community who have near falls and do not sustain an injury escape the attention of the health system. However, they are the group most likely to benefit from interventions to prevent falls. Until recently, having a fall has been the best predictor of having another fall. Recent evidence has identified clinical tests, namely single leg stance, lunge, and tandem walk five steps, that are able to discriminate near-fallers from fallers and non-fallers [6]. While gross changes in the performance of these tests are associated with falls history, there is no understanding of the contribution of postural sway to these outcomes.

Postural sway, the movement of the body over the base of support, is an indicator of balance. The traditional methods of measuring the speed, direction, and amplitude of postural sway by force plates or motion capture in gait laboratories has been superseded by wearable inertial sensors with recent interest in their measurement of standing balance [7] and gait [8]. Compared to the laboratory equipment, inertial sensors are inexpensive, portable, and permit measurements of postural sway to be taken in any setting specific to the population under investigation [9]. Additionally, wearable inertial sensors are small, lightweight, unobtrusive, and can be fixed on the body by tape, belt, or strap. Sensor data can be captured on three axes and can therefore provide detailed information in three dimensions of subtle changes in postural sway for static or dynamic conditions.

Inertial sensor measures of sway can discriminate between various age groups, and between healthy adults and adults with Parkinson's disease [10], multiple sclerosis [11], and other neurological conditions [12]. Falls risk assessment by wearable inertial sensor is more sensitive than clinical testing using the timed up and go [13]. However, the reliability and validity of inertial sensors to measure postural sway is still unclear [14], especially in seemingly healthy populations without known pathology who experience near falls and falls.

Therefore, the aim of this systematic review was to examine and synthesize the current literature on the validity and reliability of wearable inertial sensors to measure postural sway in healthy adults undertaking static and dynamic balance tests.

2. Materials and Methods

2.1. Search Strategy

Three stages of searches were undertaken, following PRISMA guidelines [15]. The first stage was to identify systematic reviews that investigated 'postural balance', 'inertial sensors', and 'reproducibility of results' via the reference list of a scoping review of systematic reviews previously conducted [16]. This search identified five systematic reviews [3,12,14,17,18]. One further relevant systematic review [7] was published after the scoping review went to press. The critical appraisal of these six recent systematic reviews [3,7,12,14,17,18] was undertaken by two independent reviewers, with a third person to mediate in the case of disagreement. None of these reviews directly answered the aims of this study. Therefore, a new search was conducted as stage two.

The existing systematic reviews assisted the development of search strategies, terms, and dates. Three main concepts informed keywords, MeSH, and search terms: 'postural control', 'inertial sensors', and 'validity/reliability' (see Appendix A for full list of search terms). Relevant truncations and expansions were applied for each database, which included Medline, PubMed, Embase, Scopus, CINAHL, and Web of Science. The dates for searching were from January 2019 to January 2021. Searches were conducted by a research librarian experienced in conducting systematic reviews.

Selection criteria followed PICO (population, intervention, comparison, outcome) principles as follows: (P) healthy adults including healthy adults as a control group; (I) wearable inertial sensor to measure static and dynamic balance; (C) force plates, motion capture or other digital or clinical measure; (O) reliability, validity, accuracy. Exclusions were for papers published with children or non-human subjects, balance or equilibrium other than postural, postural alignment, pressure sensors, and studies that investigated only static or dynamic balance, not both. Papers published before 2010 were excluded on the basis of technological advances in sensor manufacture in the past 10 years. Smartphone use was excluded because of the need to hold a device in the hand, thereby altering natural arm movement for balance maintenance or recovery [19]. Moreover, the range of balance tests interpretable by phone does not incorporate novel balance tests, such as tandem walk and lunge [6]. Only primary investigation studies were incorporated, including conference proceedings if peer reviewed. Language was limited to English.

The third search examined the reference lists of the included studies and the six systematic reviews for relevant studies that fitted the inclusion criteria.

2.2. Eligibility, Quality and Data Extraction

Two independent reviewers screened titles and abstracts against selection criteria prior to full text review. A third author was available for arbitration but was not required.

All search information was managed using Covidence systematic review software. Critical appraisal of the internal and external validity of included studies was undertaken using JBI critical appraisal checklist for analytical cross-sectional studies [20].

The first two authors extracted data from the first five studies into Excel and cross-checked for accuracy. The first author then extracted the remainder of the data, which were checked for thoroughness by the third author.

2.3. *Data Pooling*

Data pooling was multistage. Studies were initially grouped broadly to validity or reliability, then refined within these two contexts. Validity was categorized as concurrent (compared to gold standard), discriminant (able to distinguish between groups), and convergent (related to the clinical measure). Reliability was categorized as internal consistency (inertial sensor accurately measures postural sway) or test-retest reliability (sensor data replicates the results of the same postural sway activity in the same person at two time-points). Balance activities were dichotomized to static or dynamic tests, then further refined to sort into the same measurement outcomes, e.g., single leg stance for static balance; timed up and go for dynamic. Finally, the outcome measures for validity and reliability were grouped, e.g., Pearson's rho for validity; intraclass correlations for reliability. Heterogeneity was examined using $\tau 2$, I^2 and Cochran's Q statistic using the interpretations: $\tau 2 = 0$ suggests no heterogeneity, I^2 values < 25, 26–50%, and >75% suggest low, moderate, and high heterogeneity respectively, and a significant Q statistic indicated that the studies do not share similar effects [21].

2.4. *Statistical Analysis*

Interrater agreement between two reviewers was captured at three stages, namely title/abstract screen, full text inclusion, and reference list inclusion. Rater agreement was analysed using Cohen's kappa with agreement values interpreted as $\geq$0.81 excellent, 0.61–0.8 good, 0.41–06 fair and $\leq$0.4 poor [22]. For meta-analysis, homogeneity with balance activity, sensor location, and measurement outcome were required [23]. Where heterogeneity prevented meta-analysis, synthesis of the data was conducted.

3. Results

The search strategy identified 5430 articles. Following duplicate removal, as well as screening of titles, abstracts, and full text, 19 articles met the inclusion criteria. One paper repeated a previous study with different analysis and was therefore excluded [24] (see PRISMA flow diagram, Figure 1).

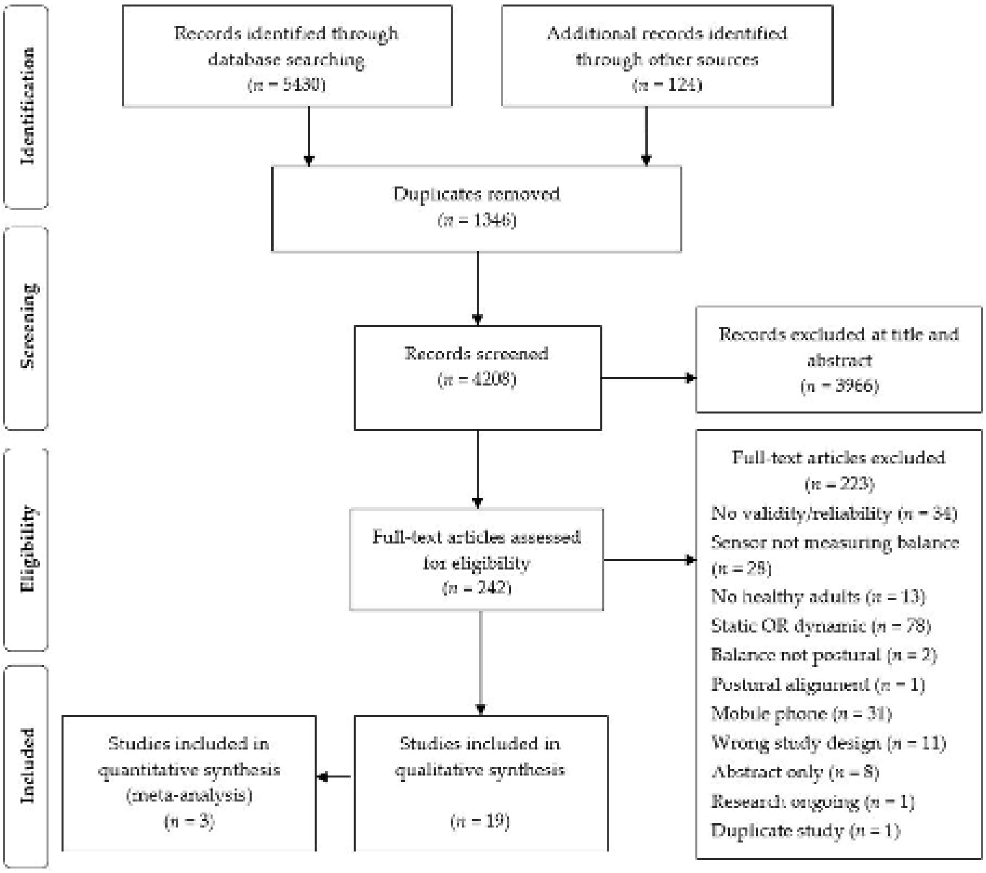


Figure 1. PRISMA flow diagram.

3.1. *Study Characteristics*

The 19 studies assessed static and dynamic balance in 1145 people, of whom 686 (59.9%) were healthy (see Table 1, Study Characteristics). Exclusively healthy populations were investigated in three studies: two in young adults [25,26] and the third in older adults [27], while healthy populations formed the control or comparison group in the remaining studies. Fallers were identified as a subject group in four studies [28–31]. The majority of papers investigated postural sway in neurological conditions, including Parkinson's disease (PD) [32–35], multiple sclerosis (MS) [36,37], Huntington's disease [38], progressive supranuclear palsy [33], muscular dystrophy [39], cerebellar ataxia [40,41], and a single case study of person with a stroke [42]. Only one musculoskeletal condition was investigated: anterior cruciate ligament reconstruction rehabilitation [43]. In Table 1, both the healthy and pathological groups are described for completeness.

Table 1. Study Characteristics.

Author, Year, Setting, Country [Reference]	Study Population, Number (Sex) Age in Years Mean ± SD (Range)	Healthy Group Number (Sex) Age in Years Mean ± SD (Range)	Static Balance Activity	Dynamic Balance Activity	Clinical Balance Measure	Outcome Measure
Brduskova et al., 2018, NA, Slovak Republic, [32]	PD *n* = 13 (8M 5F) 63.7 ± 5.7 y	Young *n* = 13 (4M 9F) 25.0 ± 2.3 y Older *n* = 13 (4M 9F) 70.1 ± 4.5 y	FA EO, FA EC	Step with vibration	NA	RMS acc AP ML, jerk, mean veloc, peak veloc, stride length, stride veloc, cadence, stance time
Craig et al., 2017, Lab, USA [36]	MS *n* = 15 (3M 12F) 48.2 ± 8.7 y	HC *n* = 15 (3M 12F) 47.8 ± 9.5 y	FA EO	7 m TUG	TUG	RMS acc ML AP V
Dalton et al., 2013, Lab, Wales, UK [38]	HD Pre-manifest *n* = 10 (4M 6F) 44.8 ± 11.7 y Manifest *n* = 14 (8M 6F) 51.8 ± 14.8 y	HC *n* = 10 (5M 5F) 56.4 ± 10.9 y	FT EO, FT EC	5 m walk	Romberg	ENMO
De Vos et al., 2020, Lab, England, UK [33]	PSP *n* = 21 (12M 9F) 71 y (63–89) PD *n* = 20 (11M 9F) 66.4 y (50–79)	HC *n* = 39 (19M 20F) 67.1 y (51–82)	FA EC	TUG, 2 min walk	TUG	min, max, mean acc AP ML V
Greene et al., 2012, Hospital clinic, Ireland [28]	Fallers *n* = 100 (NA) Whole study (57M 63F) 73.3 ± 5.8 y	Non-faller *n* = 20 (NA)	Semi TS EO 40 s, FT EC 30 s, Turn head	STS, stand to sit, transfer, fwd reach, pick up object, turn 360, place foot on stool	BBS	Peak accel, jerk, stride length, stride veloc, cadence, stance time
Hasegawa et al., 2019, NA, USA [34]	PD *n* = 144 (93M 51F) 68.4 ± 8.0 y	HC *n* = 79 (48M 31F) 68.2 ± 8.1 y	FT EO, FT EC, FT EO soft, FT EC soft, LOS, APA, APR	Step, ISAW, ISAW single task, ISAW dual task	ISAW MiniBEST	RMS acc ML AP; cadence.
Heebner et al., 2015, NA, USA [25]	NA	Healthy Reliability *n* = 10 (10M 0F) 24.3 ± 4.2 y Validity *n* = 13 (13M 0F) 24.1 ± 3.1 y	FA EO, FA EC, FAEO soft, FAEC soft, TS EO, TS EC, SLS EO, SLS EC	DPSI-AP, DPSI-ML	NA	RMS acc AP ML, mean acc AP ML, stride length, stride veloc, stance time.
Jimenez-Moreno et al., 2019, NA, England UK [39]	MD *n* = 30 (20M 10F) 48 y (25–72)	HC *n* = 14 (6M 8F) 32 y (23–47)	FA EO	6 minWT, 10 mWT, 10 m Walk/Run Test	6 minWT	Peak trunk veloc sagittal.

Table 1. *Cont.*

Author, Year, Setting, Country [Reference]	Study Population, Number (Sex) Age in Years Mean ± SD (Range)	Healthy Group Number (Sex) Age in Years Mean ± SD (Range)	Static Balance Activity	Dynamic Balance Activity	Clinical Balance Measure	Outcome Measure
Leiros-Rodriguez et al., 2016, NA, Spain [27]	NA	$n = 66$ (0M 66F) 64.9 ± 7.6 y	SLS EC, SLS EO soft	walk 10 m, turn, walk 10 m	NA	RMS acc ML AP, stride length, cadence.
Liu et al., 2012, NA, USA [29]	Fallers $n = 4$ (2M 2F) 74.5 ± 2.7 y	Young $n = 4$ (1M 3F) 21.8 y ± 1.0 y Older $n = 4$ (2M 2F) 73.3 ± 7.1 y	FA EO, FT EO, FA EC 10 s	Treadmill walk	NA	RMS acc AP ML V, jerk, sway area, path length, mean velocity, cadence.
Mancini et al., 2016, Lab (validity), clinic (reliability) USA [35]	PD Validity $n = 10$ (8M 2F) 67.2 ± 5 y Reliability $n = 17$ (12M 5F) 67.1 ± 7.0 y	HC Validity $n = 12$ (9M 3F) 68.0 ± 5.0 y Reliability $n = 17$ (6M 11F) 67.9 ± 6.0 y	FA EO	APA, first step, walk	NA	Peak acc ML AP, angular veloc, APA duration, step length, step velocity.
Martinez-Mendez et al., 2011, NA, Japan [26]	NA	$n = 10$ (7M 3F) 26 ± 3 y	FA 2 cm EO	APA, step fwd	NA	RMS acc AP ML, peak acc AP ML, sway area, jerk, trunk veloc sagittal, stride length, stride veloc, stance time, cadence.
Matsushima et al., 2015, NA, Japan [40]	SCA or CA $n = 51$ (24M 27F) 60.3 ± 10.4 y	HC $n = 56$ (28M 28F) 57.2 ± 14.1 y	FA EO, FA EC, FT EO, FT EC	walk 10 m	NA	VM horizontal acc; gait velocity, cadence, step length, step regularity, RMS ratio.
O'Brien et al., 2019, NA, USA [42]	Stroke $n = 1$ (1M 0F) 57 y	Young $n = 14$ (8M 6F) 26.4 ± 3.9 y Middle $n = 19$ (8M 11F) 43.7 ± 5.8 y Older $n = 16$ (8M 8F) 61.8 ± 5.1 y	FA EO, FA EC, FT EO, TS EO, SLS EO	10 mWT normal veloc, 10 mWT high veloc, TUG	BBS TUG	Max/mean acc AP ML V, stride length.
Rivolta et al., 2019, Rehab Centre, Italy [30]	Inpatient Fallers $n = 33$ (26M 7F) 72.7 ± 15.2 y	Inpatient $n = 46$ (30M 16F) 72.5 ± 11.5 y Volunteers $n = 11$ (0M 11F) 35.7 ± 14.0 y	FA EO, FA EC, FA EC nudge	360° turn, walk 10 m, sit to stand, stand to sit	Tinetti test	RMS acc AP ML V; mean acc AP ML V; VM; step height/length/symmetry/continuity, trunk sway.

Table 1. *Cont.*

Author, Year, Setting, Country [Reference]	Study Population, Number (Sex) Age in Years Mean ± SD (Range)	Healthy Group Number (Sex) Age in Years Mean ± SD (Range)	Static Balance Activity	Dynamic Balance Activity	Clinical Balance Measure	Outcome Measure
Senanayake et al., 2013, NA, Brunei Darussalam [43]	ACLR rehab $n = 8$ (6M 2F) 31.0 ± 4.1 y	HC $n = 4$ (3M 1F) 31.0 ± 8.3 y	SLS EO, SLS EC	Treadmill 4 kph; Treadmill 6 kph	NA	RMS acc AP ML.
Spain, St George et al., 2012, NA, USA [37]	MS $n = 31$ (12M 19F) 39.8 y (24–67)	HC $n = 28$ (9M 19F) 37.4 y (26–60)	FA EO, FA EC	T25FW, 7 m TUG	ABC, MSWS12, EDSS TUG	RMS accel AP ML, jerk, mean/peak/sway veloc, stride length, cadence, turning time, trunk rotation.
Tang et al., 2019, Uni, USA [31]	Fallers $n = 14$ Whole study $n = 30$ (13M 17F) 76.0 ± 10.5 y	Non faller $n = 16$ (NA)	FA EO	MiniBEST including TUG and dual task TUG; BBS	BBS, MiniBEST, TUG	Peak acc AP ML V, cadence, stride/step/swing, stance time.
Velazquez-Perez et al., 2020, research centre, Cuba [41]	SCA $n = 30$ (7M 23F) 43.5 ±10.5 y	HC $n = 30$ (7M 23F) 43.3 ± 10.2 y	FA EO FT, TS	10 m walk, Tandem walk 10 steps		

Key: ABC Activities-specific Balance Confidence; acc acceleration; ACL anterior cruciate ligament reconstruction; AP antero-posterior; APA anticipatory postural adjustment; APR automatic postural response; BBS Berg Balance Scale; BEST Balance Evaluation Systems Test; CA cerebellar ataxia; DPSI dynamic postural stability index (jump landing one leg); EC eyes closed; EDSS Expanded Disability Status Scale; ENMO Euclidean Norm Minus One; EO eyes open; FA feet apart; FT feet together; fwd forward; HC healthy controls; HD Huntington's Disease; ISAW instrumented stand and walk test; Lab motion analysis laboratory; m meter; min minute; MD myotonic dystrophy; ML mediolateral; MS Multiple Sclerosis; MSWS12 MS Walking Scale (12 item); NA Not available; PD Parkinson's Disease; PSP Progressive Supranuclear Palsy; Rehab rehabilitation; RMS root mean square; ROM range of motion; s second; SCA spinocerebellar ataxia; SLS single leg stance; STS sit to stand; TS tandem stance; TUG timed up and go test over 3 m; T25FW Timed 25 Foot Walk; Uni university; V vertical; veloc velocity; VM vector magnitude; WT walk test; y years of age.

Static balance activities varied by foot position (feet apart, together, tandem, semi-tandem or single leg stance), eye condition (open or closed), surface texture (firm or soft), length of time (from 10 s [25,30,39,42] to 3 min [29]), and some static activities also included perturbation (nudge or pull) [30]. Static balance was measured by the Romberg [30], Tinetti [38], or limits of stability [34] clinical tests.

Dynamic balance was assessed with postural transitions (sit to stand, stand to sit, transfer chair to chair), stepping (first step, step up), walking for a set time or distance, running, turning around, and jumping forward and sideways (Table 1). The outcome measures to evaluate these activities were varied. For example, gait was measured by step length, velocity, regularity, height, length, continuity, or symmetry; stride length or velocity; cadence and/or stance time. No single clinical test was used consistently. Dynamic clinical measures also included walking tests (timed up and go [31,33,36,37,42], 10 m walk [41], six-minute walk test [34], 25-foot walk test [37], and jumping (dynamic postural stability index (DPSI)) [25].

When both static and dynamic balance were assessed in the one balance test, they were measured by the Berg balance scale [28,31,42] and MiniBEST [31,34]. The time or distance within standardized tests differed between studies, e.g., in the TUG, the standard 3 m walking distance was increased to 7 m distance to provide more consistent data for gait parameters [36,37], and included additional single or dual tasks [31]. Static balance data from the sensors were analysed by multiple methods, most commonly root mean square (RMS) of acceleration, but also maximum, minimum, or mean of acceleration, jerk, various measures of velocity and Euclidian norm minus one (ENMO), providing challenges in grouping for meta-analysis. Dynamic balance similarly had multiple different analyses of step and stride length, stance time, cadence, and velocity.

3.2. *Quality Assessment*

All included papers were observational studies. Quality assessment used JBI analytical cross-sectional study critical appraisal checklist (see Table 2 Quality Assessment). All papers described the exposure, outcomes, and appropriate data analysis methods in detail. All studies but one [39] used standard, objective criteria. However, four studies lacked explicit selection criteria [27,29,38,39] and several provided no detail of the setting for the study [25–27,29,32,34,37,39,40,42,43], which impacts replicability. Five of the papers provided no identification or management of confounding factors [25,26,29,34,38], which impacts the trustworthiness of the results in these papers. Interrater agreement between two reviewers for screening, full text, and reference list selections was analysed using Cohen's kappa, with a result of k = 0.805 interpreted as a good result.

Table 2. Quality Assessment–JBI cross-section study.

Author, Year, [Reference]	Inclusion Criteria Defined	Subject, Setting Described	Exposure Valid Reliable	Objective Standard Criteria	Confounders Identified	Confounder Strategies	Outcomes Valid Reliable	Appropriate Stats Analysis
Bzduskova, 2018 [32]	+	-	+	+	+	+	+	+
Craig, 2017 [36]	+	+	+	+	+	+	+	+
Dalton, 2013 [38]	-	+	+	+	-	-	+	+
De Vos, 2020 [33]	+	+	+	+	+	+	+	+
Greene, 2012 [28]	+	+	+	+	+	+	+	+
Hasegawa, 2019 [34]	+	-	+	+	-	-	+	+
Heebner, 2015 [25]	+	-	+	+	-	-	+	+
Jimenez-Moreno, 2019 [39]	-	-	+	-	+	+	+	+
Leiros-Rodriguez, 2016 [27]	-	-	+	+	+	+	+	+
Liu, 2012 [29]	-	-	+	+	-	-	+	+
Mancini [1], 2016 [35]	+	+	+	+	+	+	+	+
Martinez-Mendez, 2011 [26]	+	-	+	+	-	-	+	+
Matsushima, 2015 [40]	+	-	+	+	+	+	+	+
O'Brien, 2019 [42]	+	-	+	+	+	+	+	+
Rivolta, 2019 [30]	+	+	+	+	+	+	+	+
Senanayake, 2013 [43]	+	-	+	+	+	+	+	+
Spain, 2012 [37]	+	-	+	+	+	+	+	+
Tang, 2019 [31]	+	+	+	+	+	+	+	+
Velazquez-Perez, 2020 [41]	+	+	+	+	+	+	+	+

[1] Both reliability and validity sub-studies.

3.3. *Sensors*

Sensor type, number, position, fixation, sampling frequency, and calibration methods differed between studies as outlined in Table 3. Only one study used a dual axis accelerometer (antero-posterior and mediolateral) [32] while the remainder used triaxial sensors, providing accelerometry data for the additional vertical plane. Thirteen studies also used inertial sensors with inbuilt gyroscopes providing further rotational velocity information [26,28,29,33–37,41–43]. The most common inertial sensors were Opals and XSens, where accelerometry data measured concurrent input from multiple sensors placed on the trunk and extremities. Various options for sensor body position and fixation were identified between studies. The preferred position for a sole sensor was on the lumbar spine [25,28,40,42] as this position corresponded closest to the centre of gravity of the body. Further, sensors situated on the low lumbar spine produced greater accuracy than thoracic sensors [27]. Studies using multiple sensor systems located them on the lower back, sternum, wrists, and ankles. Methods of fixation were not described in nine studies (47%). When stated, fixation from elasticated belts or bands [25,30,34–36,39,40] or adhesive tape [27,28,42] were the preferred methods. Two papers discussed movement artefacts [25,42]. However, only one excluded data due to sensor movement [42]. Sampling frequency ranged from 20 to 400 Hz, although Velazquez-Perez [41] provided no information on this. Only one paper [31] described down-sampling, which is the process of reducing the sample rate of a signal to manage the size of data. Accelerometry and gyroscopic data were analysed using sensor-specific tools [27,30], or in programs such as MATLAB [25–29,31,32,34,35,37,38,42,43] or Mobility Lab [33,34,36]. The statistical program 'R' was used in one study [39] and STASTICA in another [41].

Table 3. Sensors Overview.

Reference, Year	Sensor Type (Brand)	Number, (Body Location), Fixation	Sampling Frequency	Variables	Data Analysis Tool
Bzduskova et al., 2018 [32]	Dual axis accel (ADXL202)	2, (T4, L5), NS	100 Hz	Low pass filtered; cut-off frequency 5 Hz; Butterworth filter; calibration for ±30° range body tilt	MATLAB software
Craig et al., 2017 [36]	Triaxial accel/gyro (Opal)	6, (sternum, L5, bilat wrists, bilat ankles), elastic straps	128 Hz	Accel ranges ± 16 g, ±200 g; gyro range ±2000 deg/s	Mobility Lab software (APDM)
Dalton et al., 2013 [38]	Triaxial accel (AD-BRC)	1, (sternum), NS	250 Hz	Range ± 2.5–10 g, calibration by rotation through established angles; high pass filtered, 3rd order normalized elliptical filter, passband frequency 0.25 Hz	MATLAB software
De Vos et al., 2020 [33]	Triaxial accel/gyro (Opal)	6, (sternum, L5, bilat wrists, bilat feet), NS	100 Hz	Wireless data stream to laptop	Mobility Lab software
Greene et al., 2012 [28]	Triaxial accel/gyro (SHIMMER)	1, (L3), adhesive tape	102.4 Hz	Calibration using standard method; data streamed via Bluetooth to laptop	MATLAB
Hasegawa et al., 2019 [34]	Triaxial accel/gyro (Opal)	8, (sternum, L5, bilat wrists, bilat shins, bilat feet), elastic straps	128 Hz	Unscented Kalman Filter	Mobility Lab (APDM) and MATLAB
Heebner et al., 2015 [25]	Triaxial accel (ADXL78)	1, (L5), neoprene belt	100 Hz	Range ± 16 g, built in data acquisition and storage, low pass filter 50 Hz	MATLAB
Jiménez-Moreno et al., 2019 [39]	Triaxial accel (GENEActiv)	4, (bilat wrists, bilat ankles), elastic band	100 Hz	Output metric ENMO–mg.	R software
Leirós-Rodríguez et al., 2016 [27]	Triaxial accel (GT3 Plus)	3, (T4, L4, L5), adhesive tape	100 Hz	Configured 1 s timeframe. Concurrent analysis video & accelerometry data; reviewed analysis.	ActiLife software
Liu et al., 2012 [29]	Triaxial accel/gyro (MTX Xsens)	2, (L5, ankle), NS	50 Hz	Maximum Lyapunov exponent	MATLAB
Mancini et al., 2016 [35]	Triaxial accel/gyro (Opal validity; MTX Xsens reliability)	6 validity/3 reliability (sternum, L5, bilat wrists, bilat ankles) elastic straps	128 Hz Opal; 50 Hz MTX Xsens	3.5 Hz cut-off, zero-phase, low-pass Butterworth filter. Resampling from inertial sensor, force platform and infrared cameras at 50 Hz.	MATLAB

Table 3. *Cont.*

Reference, Year	Sensor Type (Brand)	Number, (Body Location), Fixation	Sampling Frequency	Variables	Data Analysis Tool
Martinez-Mendez et al., 2011 [26]	Unit with triaxial accel (MMA, Freescale) & gyros (X3500 Epson; ENC-03RC Matura)	2, (L3/4, ankle of dominant foot), NS	100 Hz	Accel range ± 1.5 g, gyro range ± 80 deg/s; response freq 0.01–58 Hz. Bluetooth transmission	MATLAB
Matsushima et al., 2015 [40]	Triaxial accel (Jukudai Mate)	1, (L3), elastic belt	20 Hz	Detection range ± 10 g; resolution power 0.02 g	BIMUTAS II
O'Brien et al., 2019 [42]	Triaxial accel/gyro (BioStampRC)	1, (L5), Tegaderm adhesive film	31.25 Hz	Accel ± 4 g; gyro ± 2000 deg/s; 4th order low pass Butterworth filter 2 Hz; acquisition with BioStampRC	MATLAB
Rivolta et al., 2019 [30]	Triaxial accel (GENEActiv)	1, (chest), elastic band	50 Hz	12 bits over range ± 8 g; chronometer for starting time; high pass 3rd order Butterworth filter	Manually segmented accel signals; GENEActiv software
Senanayake et al., 2013 [43]	Triaxial accel/gyro (KinetiSense)	4, (bilat thighs, bilat shins), NS	128 Hz	Wireless transmission via USB	KinetiSense and MATLAB
Spain, St George et al., 2012 [37]	Triaxial accel/gyro (XSens)	6, (sternum, L5, bilat wrists, bilat ankles), NS	50 Hz	Accel range ± 1.7 g; gyro range ± 300 deg/s. Filtered with 3.5 Hz cutoff, zero phase, low pass Butterworth filter	MATLAB
Tang et al., 2019 [31]	Triaxial accel (ADXL330)	2, (hip, foot), NS	400 Hz, down sampled to 25 Hz	Common and Activity Specific features extracted; mRMR feature selection	MATLAB
Velazquez-Perez et al., 2020 [41]	Triaxial accel/gyro (Opal)	6, (Hands, feet, sternum, L5), NS	NS	NS	STATISTICA

Key: accel accelerometer; deg degrees rotation; freq frequency; g gravitational velocity (m/s^2); gyro gyroscope; Hz hertz; mRMR minimal-redundancy-maximal-relevance; NS not stated; s second.

3.4. Validity

The validity of the inertial sensor to measure balance was explored through concurrent (compared to gold standard), discriminant (able to distinguish between groups), and convergent (related to the clinical measure) validity. Data pooling was not possible for meta-analysis concerning validity due to the variety of protocols and outcome measures undertaken.

Concurrent validity was assessed by comparing the inertial sensor with force plates for static balance in six studies [25,26,28,29,32,35] and with force plates or motion capture systems for dynamic balance in three studies [35,38,42]. The resultant correlations identified that inertial sensors provide moderate to strong evidence of concurrent validity for medio-lateral (ML) (r = 0.58–0.84) [25,28,30] and antero-posterior (AP) sway (r = 0.71) [26] in static balance. There were good to excellent correlations between inertial sensor and instrumented walkway for step time (ICC 0.68–0.92), step length (ICC 0.68–0.89), and gait velocity (ICC 0.90–0.94) [25,38,42].

Discriminant validity was used to compare inertial sensor measures between young and older healthy participants [29,32,42], fallers and non-fallers [28–31], and healthy controls from people with specific diagnosed conditions [32–40,43]. Young adults showed significantly less medio-lateral sway than older adults during static stance [32,42]. The same was true for dynamic activities including gait velocity, step length, turning speed and stand to sit [42]. Inertial sensors were able to distinguish sway differences between fallers and non-fallers [28–31]. Dynamic balance activities to discriminate fallers from non-fallers included AP acceleration of walking [29,30], and functional activities of stepping on a stool and sitting to standing [30,31]. Inertial sensor classification accuracy for discriminating fallers from non-fallers ranged between 72.24% (95%CI 69.84–74.52%) [28] and 89% [30]. Compared to diagnostic populations, inertial sensors discriminated healthy controls by significantly reduced sway amplitude in eyes closed condition [32–34,37,40,42] and increased anticipatory postural adjustments [32,34,35,38,39] during standing. In dynamic balance, both walking cadence and turning velocity discriminated healthy controls [33,34,37]. Neither sit to stand nor stand to sit activities discriminated diagnostic groups from healthy controls. Discrimination accuracy varied between studies. There were moderate to strong results to discriminate healthy controls by sway acceleration amplitude in standing (AUC 0.68) [37], with eyes open or closed (classification accuracy 94–96%) [43] and lateral trunk range of motion in gait (AUC 0.72) [37].

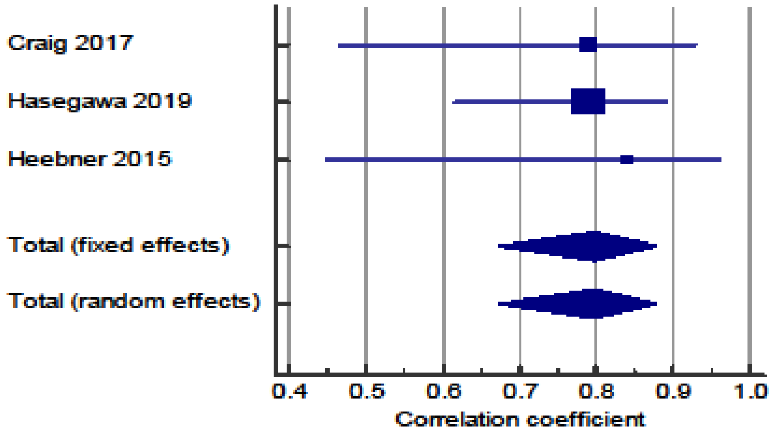
Regarding sensor position, a single lumbar spine sensor identified significant differences between younger and older healthy adults [32,42] and was able to distinguish the different dynamic balance tasks of lateral and forward jumps [25]. Further, the single sensor was as accurate as the six-sensor array [33].

Convergent validity evaluated the inertial sensor balance measures against clinical balance tools. The six-sensor array identified differences between the study group and healthy controls when observed, whereas timed clinical tests could not [33,37].

3.5. Reliability

Reliability was investigated in eight papers [25–27,29,34–36,39]. Internal consistency was assessed in three studies by evaluating test results across multiple sensors during the same activity (ICC 0.62 to 0.98) [26,27,39]. Inter-accelerometer reliability was good between right and left limbs (ICC > 0.8) [39], between L4 and L5 (r = 0.78–0.95) [27], between thoracic and lumbar spine (r = 0.60–0.76) [27] and between lumbar spine and ankle (inter-item correlation 0.70–0.98) [26], but not between upper and lower limbs (ICC = 0.59) [39].

Test-retest reliability showed reasonable consistency between studies. Meta-analysis was possible when static stance incorporated feet apart eyes open, measured by RMS of acceleration ML and AP, and when intraclass correlations were undertaken for statistical analysis [25,34,36]. Results from the grouped studies produced high homogeneity ($I^2 = 0.0\%$) with similar effects (Cochrane's Q non-significant 0.14) indicating trustworthiness of the sensors to measure static balance (Figure 2). However, the lower quality of two of the included papers [25,34] (Table 2) influenced the strength of findings. Therefore, meta-analysis results were considered informative rather than conclusive. Measurements of static sway distance, sway area, path length, mean velocity, and RMS were reliable, indicated by moderate to good correlations ranging between ICC 0.57 and 0.79 [34,36]. Although dynamic balance was measured in diverse balance tasks, all test-retest parameters of dynamic balance produced moderate to excellent correlations (ICC 0.696–0.94), indicating strong correlations and good reliability in healthy adults [25,34,36].



Cochrane's Q	0.1412
DF	2
Significance level	$p = 0.9318$
I^2 (inconsistency)	0.00%
95% CI for I^2	0.00 to 52.48

Figure 2. Meta-analysis.

4. Discussion

The aim of this systematic review was to investigate and synthesize the validity and reliability of wearable inertial sensors to measure postural sway in static and dynamic balance for healthy adults. Test-retest reliability results were consistently moderate to excellent for static and dynamic balance across the included studies. Meta-analysis was impossible for the validity studies due to heterogenous samples and methods. However, the synthesis showed moderate to good validity overall. These findings indicate consistency against gold standard equipment for measures of ML and AP sway in static balance and step time, step length, and gait velocity for dynamic balance. While the sensors were able to discriminate young from old, and fallers from non-fallers, the accuracy of discriminating healthy controls from diagnostic groups varied between studies.

The variability in equipment included multiple types of sensor. While all studies used accelerometer data, fewer included gyroscope data, suggesting data from accelerometers may be sufficient for clinical interventions. This reduced complexity may encourage more clinicians who are unfamiliar with the technical aspects of the new equipment to integrate sensors into practice. The multiple strategies for data acquisition, feature extraction, signal processing, and data analysis presented a heterogenous mix unsuitable for meta-analysis.

There was no consistent number of sensors or sensor placement position. However, the lumbar spine (L3–L5) was the preferred site overall. A single inertial sensor was as reliable as multiple sensors when placed near the centre of mass (L3–L5) and showed moderate to good validity and test-retest reliability for both static and dynamic balance. A single sensor placed over the centre of mass would provide simplicity in the clinical setting, particularly during telehealth interactions when instruction, observations, and interventions are provided remotely. A single sensor also aligns with recent literature for identifying differences between fallers and non-fallers [44]. However, using different placements for static and dynamic balance activities [29], and different body positions for sensor fixation, created challenges with pooling data. While different research questions demand different types of analysis, the standardization of the sensor position would permit a comparison of results across studies.

The validity of sway measures from wearable inertial sensors compared to the gold standard force plates or motion capture provided promising results across studies. These results concur with a previous scoping review of systematic reviews [16] as well as recent studies investigating the concurrent validity of sensors to measure balance in healthy adults [8,45–47]. In healthy populations, this indicates that inertial sensors provide valid data when used in home and community settings [48]. This provides flexibility for clinical treatment and trials, particularly in rural and remote settings, or during social distancing such as with COVID-19 [49]. Importantly it ensures that performance during testing is not altered by an unfamiliar environment. Therefore, these findings provide reassurance that the sensors are a valid proxy for the gold standard as a means of measuring static and dynamic balance in the community.

Sensors were valid in discriminating sway between younger and older participants, reinforcing the sway changes that occur due to ageing [50]. Sensors also discriminated fallers from non-fallers. The sensor data discriminated sub-tasks within clinical tests such as separating components for the timed up and go into the sit-to-stand, walk straight, turn, and stand-to-sit, which is consistent with previous findings in timed up and go [51]. While several studies measured the sway differences between fallers and non-fallers, no studies investigated differences between non-faller, fallers, and those who had experienced near falls. As this review provides evidence that sensors can identify subtle changes in sway between different aged healthy people, it is possible that sensors may identify sway differences between near fallers, fallers, and non-fallers. The early detection of subtle changes in postural sway is required to identify the risk of near falls [4,52] and can be measured reliably and with confidence of validity using inertial sensors.

The main limitation to this investigation was the inability to pool included studies for meta-analysis due to heterogeneity with balance activity, sensor location, and measurement outcomes. Additionally, some limitation may be considered from the inclusion of articles written only in English.

5. Conclusions

Measuring postural sway using inertial sensors in healthy adults permits assessment and treatment in the person's natural environment, providing reassurance of accurate measures during times of social distancing. The ability to identify separate components of clinical tests using sensors permits the detection of subtle sway changes that may contribute to understanding sway differences for near falls as well as falls. Further research is required to evaluate the convergent validity of using a single sensor over the centre of mass rather than a six-sensor array for clinical balance tests such as the timed up and go test. Similarly, further research using a single sensor to discriminate sway differences between healthy and diagnostic groups, distinct age groups, and fallers/non-fallers would encourage the clinical uptake of sensors.

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Appendix A

Table A1. Search strategy for Ovid Medline.

#	Searches	Results
1	postural balance/or posture/or standing position/	81,783
2	(balance* or postur* or sway* or stability or equilibrium).ti,ab,kf.	854,996
3	("center of pressure" or "centre of pressure").ti,ab,kf.	4573
4	(stumbl* or near* fall* or misstep* or mis step*).ti,ab,kf.	1635
5	or/1–4 [Balance concept]	900,273
6	mobile applications/or cell phone/or smartphone/	15,528
7	Accelerometry/or Magnetometry/	5247
8	((body or motion or wearable*) adj2 sensor*1).ti,ab,kf.	4717
9	(acceleromet* or gyroscop* or magnetomet* or goniomet* or inclinomet* or baromet*).ti,ab,kf.	28,201
10	or/6–9	84,193
11	"reproducibility of results"/or "sensitivity and specificity"/or "predictive value of tests"/	774,004
12	(accura* or assessment* or measur* or evaluat* or reliab* or reproduc* or consistenc* or repeatab* or validit* or sensitiv* or specificity or respons* or clinimetric or correlat* or concord* or discrim*).ti,ab,kf.	11,023,968
13	or/11–12	11,173,078
14	and/5, 10, 13	4689
15	limit 14 to (english language and yr = "2019–Current")	711

Database: Ovid MEDLINE(R) and Epub Ahead of Print, In-Process & Other Non-Indexed Citations and Daily 1946 to 11 January 2021.

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C Manuscript: Balance Provocation Tests Identify Near Falls

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BALANCE PROVOCATION TESTS IDENTIFY NEAR FALLS IN COMMUNITY ADULTS AGED 40-75 YEARS: AN OBSERVATIONAL STUDY.

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Author's note

Prior Presentation of material

The data reported in this paper is a subset of a large dataset collected to establish a profile of healthy ageing in independent Australian community-dwelling adults aged 40-75 years (published 2019). The odds ratio results were presented orally by the primary author at the Australasian Association of Gerontology conference in November 2020.

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Conflict of Interest

The authors declare that they have no competing interests.

Author contribution

Nicky Baker: conceptualisation, methodology, investigation, data curation, data analysis, writing original draft, review and editing

Karen Grimmer: conceptualisation, validation, data curation, formal analysis, writing review and editing

Susan Gordon: conceptualisation, methodology, investigation, data curation, resources, supervision, funding acquisition, project administration, writing review and editing

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Balance provocation tests identify near falls in healthy community adults aged 40-75 years; an observational study.

ABSTRACT

Background: Near falls, such as stumbles or slips without falling to the ground, are more common than falls and often lead to a fall.

Purpose: The objective of this study was to investigate which balance tests differentiate near fallers from fallers and non-fallers.

Methods: This cross-sectional, observational study assessed balance in healthy community dwelling adults aged 40-75 years. Participants reported falls and near falls in the previous six months. Balance testing was completed in local community for static (feet together and single leg stance) and dynamic balance (tandem walk, Functional Movement Screen hurdle step and lunge). Between-group comparative analysis of pass-fail for each balance test was undertaken.

Results: Of 627 participants, there were 99 fallers (15.8%), 121 near fallers (19.3%) and 407 non-fallers (64.9%). Near fallers were twice as likely as non-fallers to fail single leg stance eyes (OR 2.7, 95% CI 1.5-4.9), five tandem steps (OR 2.5, 95% CI 1.5-5.7), hurdle step (OR 2.9, 95% CI 1.4-5.8) and lunge (OR 2.5. 95% CI 1.5-4.1). The predictive capacity differentiates near fallers with sensitivity of 73.3%.

Discussion: A new battery of tests assessing static and dynamic balance identify near fallers in seemingly healthy, community dwelling middle- and young-older age adults.

(Word count 200 words)

Keywords: **postural balance, falls, near falls, methods**

INTRODUCTION

A fall is an unanticipated loss of balance where the person lands on the ground or lower surface (WHO, n.d.). Current practice for falls risk is targeted at known fallers (Steffen et al., 2002) or those with known conditions that increase risk of falling. Little is known about near fall events during the pre-fall phase. Near fall events are those moments of balance loss that do not conclude in landing on the ground and can be defined as "*slips, trips, stumbles, missteps, incorrect weight transfer, or temporary loss of balance*" (Pang et al, 2019). Near-fall events are more common than falls and often pre-empt a fall (Nagai et al, 2017). They commonly occur when seemingly healthy people trip or stumble but disregard the event because no fall or injury occurs. Identifying people in the near fall phase provides an opportunity to inform clinical practice to address mitigating factors and decrease falls risk.

Current falls risk screening is well established after a calamitous health event such as hip fracture (Karantana et al, 2011) or stroke (Xu et al, 2018) and in neurological conditions such as Parkinson's Disease (Caetano et al, 2018) or multiple sclerosis (Quinn, Comber, Galvin, and Coote, 2018). Falls risk screening is also readily available for older adult community dwellers through clinical functional movement tests such as the Timed Up and Go, Berg Balance Scale and Five Time Sit to Stand (Lusardi et al, 2017). Risk factors for falls in middle aged adults have identified a reduction in body position sense and postural control (Verma et al, 2016), reduced functional mobility and poor bladder control (Peeters et al, 2019) as well as diminished vision, obesity, poor muscle strength (White et al, 2018) and comorbidities of diabetes and osteoarthritis (Singh et al, 2019). Falls screening activities in the middle-age group include physical performance measures of standing on soft surface with eyes closed (Bareis et al., 2018), reaction time and proprioception (Pang et al, 2021). Near-fall risk screening has been investigated in specific populations such as stroke (Gangwani et al, 2020) and Multiple Sclerosis (Fritz et al, 2018) and in healthy populations by using foot pressure sensors in a laboratory setting (Niu et al, 2019).

For community population testing, balance tests need to be safe, appropriate for a target population with variable levels of fitness or mobility, readily and efficiently administered, sensitive to balance constraints, not require sophisticated equipment, and be able to provide population norms (Langley and Mackintosh, 2007). However, no accepted set of balance assessments exist to differentiate between non-fallers, fallers and people experiencing near fall events (Noohu, Dey, and Hussain, 2014), nor the balance of people aged under 65 years without known balance problems. This paper investigates the balance performance, its association with falls, near falls, age and gender, and the reasons for any falls or near falls, in seemingly healthy, community dwelling people aged 40 to 75 years. This study aimed to test the predictive capacity of static and dynamic balance tests in differentiating between fallers, near fallers and non-fallers.

METHOD

Design

The design was cross-sectional and observational. Data reported in this project were a subset of a larger study to establish a profile of healthy ageing in independent community-dwelling adults aged 40-75 years ([redacted]). Participant selection criteria, recruitment and provision of information on physical, social, emotional and mental health via survey responses and physical testing are described previously ([redacted]). Ethical approval was provided by the [redacted] Clinical Human Research Ethics Committee (391.16).

Subjects

Parent study participants were seemingly well, community dwelling adults aged 40-75 years who were recruited for healthy ageing general screening from local business and council networks. Falls, near falls, balance and functional fitness were components of the general health screening. Participants each provided informed, written consent, and additional verbal consent prior to each individual test. Participants attended seven stations in total, one station at a time, but not in strict number order ([redacted]). Each participant's balance was assessed by health students under the supervision of experienced, registered health clinicians ([redacted]). Student training occurred prior to all assessments and was reinforced on the day by online training videos and supervisor feedback. Interrater reliability was not assessed.

Participants responded to "Have you had a fall to the ground in the previous six months?" and "Have you had a near fall such as a stumble, slip or trip and recovered your balance, in the previous six months?" with yes/no. The number and mechanism of each fall or near fall event was recorded and near fall all responses were taken at face value. Participants who reported falling to the ground were categorised as a 'faller' irrespective of near fall events; participants who reported a near fall but no fall to the ground were categorised as 'near fallers'; and participants who reported neither fall nor near fall were categorised as 'non-fallers'. Categories were mutually exclusive.

Procedure

As no validated method has been reported to test balance in seemingly well, community-dwelling adults aged from 40 years (Pardasaney et al, 2012), a new combination of balance tests was developed consisting of ‘core’ and ‘additional’ tests (see Appendix 1). The core tests had been validated to identify fall risk in older adults (feet together eyes open 30 seconds, feet together eyes closed 30 seconds, single leg stance eyes open 5 seconds, turn 180 degrees, tandem walk forwards five steps) (Mackintosh, Datson, and Fryer, 2006). The ‘additional’ tests were designed to challenge balance in this younger, healthy population by reducing visual input, holding the position for longer and incorporating functional activities. The additional tests consisted of single leg stance eyes open for 30 seconds, single leg stance eyes closed for 5 seconds, and if this was managed then also for 30 seconds (Springer et al, 2007) and tandem walk backwards (Elboim-Gabyzon and Rotchild, 2017). The Functional Movement Screen (FMS) assessed functional balance with the hurdle step and in-line lunge (Cook, Burton, and Hoogenboom, 2006). Due to the population testing safety aspect, the lunge was completed initially on the floor and if the participant was successful, then attempted on the FMS narrow beam. Final scoring of each test was pass or fail. Right and left sides were initially scored independently, then the lower score used as the overall score for that test (Perry and Koehle, 2013). Activity scoring for the hurdle step and lunge was based on the published protocol (Cook, Burton, and Hoogenboom, 2006). In this study, an FMS score of 0 or 1 constituted a fail and an FMS score 2 or 3 constituted a pass. When a participant declined to attempt, or attempted and failed, a balance test, subsequent additional tasks on that limb were not permitted for safety, resulting in fewer participants attempting additional, progressively more difficult tasks. The random order of testing in the parent study moderated fatigue levels affecting balance performance. Further, fatigue was mitigated by gaining verbal consent before each activity and providing a rest area with refreshments that participants could access freely throughout the testing session.

Data Analysis

Data were analysed per protocol for the 627 participants in this study. Intention to treat with last value carried forward was analysed to preserve power and assess result accuracy (see Appendix 2). Associations between group and age were calculated with independent samples Kruskal-Wallis test. Group, gender and balance test result (pass/fail) associations were calculated with Pearson chi-square tests for independence with effect size (ES) calculated for two degrees of freedom with Cramer’s V (small = 0.07, medium = 0.21, large = 0.35) (Pallant, 2020). Significance was set at $p < 0.05$. Odds Ratios (ORs), sensitivity, specificity, positive and negative predictive values of each test were calculated between groups to estimate relative risk of failing the balance tests. The 95% confidence interval (CI) was considered significant when it did not contain the value 1.0. An OR of >1 was positively related to failing the balance test and conversely an OR of <1 was considered protective of failing. When numbers in any cell were less than 5, ORs were not calculated (Portney and Watkins, 2014).

Sensitivity and specificity measures were calculated to establish validity of the tests. Results closer to 100% were considered stronger prediction (Portney and Watkins, 2014). Sensitivity testing identified true positives, fallers or near fallers, who failed tests compared to non-fallers. Specificity testing identified true negatives (non-fallers) who were likely to pass the tests. Fall or near fall were compared against near fall or non-fall, respectively. Similarly, the positive predictive value was interpreted as the probability that the person would have had a fall or near fall when the test was positive (failed) and negative predictive value was considered the probability that the person had not had a fall or near fall when the test result was a pass.

For each test, failure to pass the test was scored as 1. Data distributions were determined for each composite score, and normality was tested using Shapiro-Wilks tests, where $p<0.05$ indicated non-normal distributions. For clinical application, the balance tests that significantly differentiated between fall groups were combined into one predictor variable to investigate if failing one, two or more tests would identify near fallers. The variables were combined by two methods:

- Ordinal: Summed binary scores for each test. This score had a minimum of 0 (passed all tests) and a maximum of six (no tests passed) per person.
- Factor analysis: Summing the individual test loadings produced a weighted total score. Factor analysis condenses multiple precursor variables to identify latent variables that may not be measured directly (Child, 2006). Principal component analysis and varimax rotations confirmed the weightings and arrangements of latent factors; important factor components were weighted ≥0·30. Factors with the highest weighting variable were retained; however, where a variable had similar weightings across more than one factor, decisions regarding its best placement were made on *a priori* clinical basis. Per-participant scores for each factor were calculated by multiplying their at-risk score for each variable (0 or 1, as described above) by the factor weighting, then summing the factor weightings. Therefore, if a participant passed a test, the contribution of that variable to their overall score was zero (0*loading); and conversely, if a participant failed that test, the variable contribution to the total score was 1*loading (Polit and Beck, 2012).

Sensitivity analysis: Receiver Operator Characteristic (ROC) curves were calculated to compare predictive capacity of the composite scores for consecutive groups (non-faller and near faller; near faller and faller) (Fawcett, 2006). As there is no robust information on predictors of near falls in Australian community-dwellers aged 40-75 years, we assumed an unknown ratio of positive and negative cases. The ROC curves tested the predictive capacity of composite scores to differentiate between groups. The findings were reported as the Youden Index, (a summary measure of the predictive capacity of the ROC curve), sensitivity, specificity, criterion value (best cut point trade-off) and area under the curve (AUC (95%CI)). AUC was significant if the lower 95%CI did not include 0.5 (an indicator of no predictive capacity). Statistical analyses were conducted in MedCalc, SAS and SPSS.

The reasons provided for falls and near falls were analysed by two of the research team and categorised to two themes: intrinsic and extrinsic. 'Intrinsic' related the cause of fall or near fall from within the body such as cardiovascular, sensory or musculoskeletal systems while 'extrinsic' reasons were considered external to the body, such as environment or clothing.

RESULTS

Of 656 participants in the parent study 29 were excluded. Twelve provided no information on fall or near fall history therefore could not be allocated to a group; 17 did not undertake any balance activities. This left 627 people (95.6% parent cohort) who undertook balance testing: 407 (64.9%) non-fallers (mean age 59.5 +/- 10.6 years, 64.0% female); 121 (19.3%) near fallers (mean age 60.7 +/- 9.6 years, 72.9% female) and 99 (15.8%) fallers (mean age 61.6 +/- 11.2 years, 73.7% female). No adverse events occurred during the balance assessments. There was no significant association between fall group and age (χ^2 (2, n = 627) = 4.18, p = 0.12) or gender (χ^2 = 5.60, p = 0.06, Cramer's V = 0.06).

Intrinsic factors subcategorised to the cardiovascular system (n = 10) e.g., 'blackout'; the musculoskeletal system (n = 24) e.g. 'weak hip gave way'; cognitive load (n = 13) e.g. 'distracted'; or other sensory reason (n = 34) e.g. 'leg went numb'. Extrinsic reasons related to the environment, such as uneven surfaces (n = 99) e.g., 'tripped on tree roots'; wet surfaces (n = 22) e.g. 'slip on tiles'; low light (n = 5) e.g. 'fell off step in dark'; unexpected external perturbation (n = 17) e.g. 'on boat, rough weather'; or from ill-fitting attire (n = 11) e.g. 'pants too long'. The most frequent reason for fall (n = 36) and near fall (n = 63) events was tripping. Of the fallers, the majority reported a single fall in the previous six months. However, seven fallers reported two falls (7.1%), none reported more than two, and 35 fallers (35.4%) also reported near falls. Seven of the near fallers (5.8%) reported more than one near fall but no falls.

Significant between-group differences with moderate effect size were identified for single leg stance eyes open for five seconds (SLSEO5s), turn 180 degrees (Turn180), tandem steps forward five steps (TFwd), the FMS hurdle step (HS), lunge on the floor (LunFl) and FMS lunge on the beam (LunBe). No significant association was found between the near fallers and fallers for any of the balance tests (see Table 1).

<< insert Table 1 about here >>

Differences between fallers, near fallers and non-fallers were identified by odds ratio (OR) calculations (see Table 1). Near fallers were more than twice as likely to be unsuccessful as non-fallers for the same tests: SLSEO5s, TFwd, HS, LunFl and LunBe. These tests, plus turning 180 degrees, were significantly associated between fallers and non-fallers. Further, sensitivity testing for individual balance tests identified a positive (failed) test for fallers or near fallers in SLS eyes closed (EC) for 30 seconds. Specificity testing identified negative (passed) tests for non-fallers in SLSEO5s, tandem walk forwards and backwards, hurdle step and the floor lunge (Table 1). When intention to treat analysis was conducted, there were no significant associations between fallers and near fallers, as per protocol analysis. Similarly, it produced significant associations and Odds Ratios in the same tests as identified in protocol analysis between fallers and non-fallers, and near fallers and non-fallers (Appendix 2). The predictive capacity of the tests to identify risk of near fall was evaluated by combining test result scores.

The ordinal composite score was non-normally distributed (Shapiro-Wilks test 0.91 ($p<0.001$). The overall median (IQR) was 1 (0-2) (range 0-5) and the different groups were no fall 1 (0-2); near-fall 2 (1-3); and fall 2 (0-3). There was significant between-score difference across the three groups (Wilcoxon Rank test 11.3 (df=2) $p<0.001$). Comparing consecutive paired fall groups, the significant difference remained between non-fallers and near fallers (Wilcoxon Rank test 7.5 (df=1) $p<0.05$); however, there was no difference between near fallers and fallers (Wilcoxon Rank test 0.5 (df=1) $p>0.05$). Non-fallers compared with near fallers had modest predictive capacity (Area Under the Curve (AUC) 0.61 (0.56-0.65) $p<0.05$). The threshold value was >1, sensitivity 72.7%, specificity 49.4%) (See Figure 1A). Near fall compared with fall had poor predictive capacity (AUC 0.51 (0.44 to 0.58) $p>0.05$). The threshold value was $\leq$1, sensitivity 38.4%, specificity 72.7% (Figure 1B).

<< Insert Figure 1 about here >>

Factor-weighted composite score: Only one factor was identified. The test weightings used to develop a composite score were (0.59*SLSEO5s) + (0.43*Turn180) + (0.59*TFwd) + (0.67* HS) + (0.77*LunFl) + (0.77*LunBe). This score was not normally distributed (Shapiro-Wilks test 0.89 ($p<0.001$)). The overall median (IQR) was 1.44 (0.43-2.20) (range 0-3.39) and the three groups were non-fallers 1.26 (0-2.23); near fallers 1.54 (0.67-2.21); and fallers 1.54 (0.59-2.21). As with the ordinal score, there was a significant between-group median score difference (Wilcoxon Rank score 20.6(df=2) $p<0.05$). Comparing consecutive paired groups, the significant difference remained between non-fallers and near fallers (Wilcoxon Rank test 31.7(df=1) $p<0.05$), but not between near fallers and fallers (Wilcoxon Rank test 0.9 (df=1) $p>0.05$). The predictive capacity of non-fallers compared with the near fallers was modest (AUC 0.61 (0.57 to 0.67) $p<0.05$). The threshold value was >1.02, sensitivity 73.3%, specificity 49.7% (Figure 1C). Near fallers compared with fallers had poor predictive capacity (AUC 0.51 (0.44 to 0.57) $p>0.05$). The threshold value was ≤0.77, sensitivity 38.4%, specificity 72.7%. (Figure 1D).

In summary, both ordinal and factor scores were equivalent at differentiating between non-fallers and near fallers, and between the near fallers and fallers. Failing two or more tests was predictive of a near fall.

DISCUSSION

This study investigated the association between fall group allocation and balance performance tests, to differentiate between non-fallers, near fallers and fallers in middle- and young older-aged adults. Tests to distinguish near fallers from non-fallers are important because although neither group has yet fallen, near fallers are more likely to fall in the future (Nagai et al, 2017). These results provided new information on the balance abilities of the near fall group who had statistically significant, strong association with the likelihood of failing key static and dynamic balance tests compared to their non-faller counterparts. These balance activities assessed postural control by testing functional mobility, core stability, and lower limb strength and coordination. Problems with these have already been identified as fall risk factors in middle age (Peeters et al, 2019; Verma et al, 2016; White et al, 2018). The inability of near fallers to complete these tests compared to non-fallers provides further information regarding future falls risk in this age group. Further, physiotherapists are ideally placed to incorporate functional mobility, core stability and lower limb strength and coordination training at any stage of rehabilitation or therapy. This would mitigate any such deficits and reduce the risk of future near falls or falls.

Over a decade ago, it was identified that older adults who sustained multiple near falls were more than three times as likely to sustain a future fall (Srygley, Herman, Giladi, and Hausdorff, 2009). The odds ratios in this study are comparable for the younger aged cohort, identifying near fallers as more than two times more likely to fail the tests (Table 1), than the non-fallers. This aligns with previous research on older adults identified near falls as significant predictors for an upcoming fall (hazard ratios of 5.5 – 7.6) (Nagai et al, 2017). In that prospective study, 25% of participants experienced near falls within the month of testing, whereas only 19% of participants reported near falls retrospectively in this study. Rapp et al. (2014) found slightly fewer incidents were reported retrospectively compared to prospectively, which aligns with these findings (Rapp et al, 2014).

The results of this study provide a battery of balance tests to distinguish middle and young-older aged people at risk of a near fall, permitting early intervention to avoid a future near fall or fall event (Peeters et al, 2018). The opportunity to identify balance parameters while prevention is still an option is important to population health (Gale, Westbury, Cooper, and Dennison, 2018). The results of this study confirmed that known fallers were more likely than non-fallers to fail static and dynamic balance tests, aligning with previous clinical testing protocols such as Timed Up and Go (Batko-Szwaczka et al, 2020) and Berg Balance Scale (Menezes et al, 2020). This new cohort of tests showed a significant association to distinguish near fall from non-fall participants, thereby providing an opportunity to identify adults with covert functional decline at risk of future near fall. Limitation with self-reporting falls or near falls is well-established, whether data are gathered retrospectively or prospectively (Kunkel, Pickering, and Ashburn, 2011). This limitation will remain until falls and near falls are recorded objectively such as with smartphones (Pang et al, 2019; Roeing, Hsieh, and Sosnoff, 2017). However, for this cross-sectional study design, retrospective recall, supported by prompts regarding number and cause, was considered the best available method to record falls or near falls (Rapp et al, 2014).

This study identified static and dynamic balance tests that discriminated between near fall and non-fallers, identifying risk of near falls by 73%. Further investigation has commenced to establish objective differences between these groups, based on this set of tests.

IMPLICATIONS FOR PHYSIOTHERAPY PRACTICE

The balance tests are safe, applicable in clinical and population screening activities, and use readily available equipment. Failing two or more of the tests indicates near fall risk. Given the ease of measuring single leg stance for 5 seconds, stepping into a lunge on the floor and walking forward for five tandem steps in the clinical setting, these tests could be selected to identify near fall risk in the healthy middle and young-older population who have not yet fallen. Clinically, the identification of adults with covert functional decline at risk of a near fall permits early intervention and future fall prevention.

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Figures and Tables

Figure 1 Composite score predictive ability

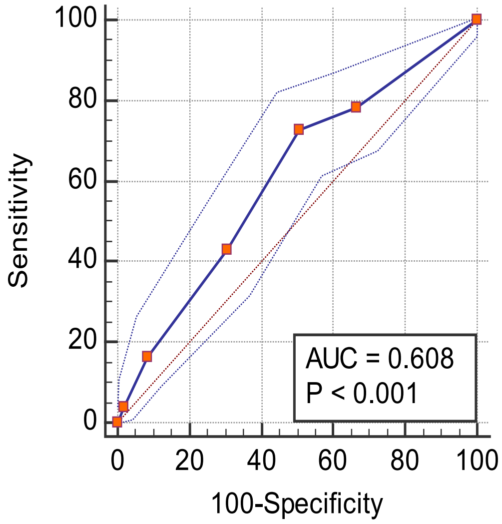

Figure 1A Ordinal composite score Near Falls versus Non-faller

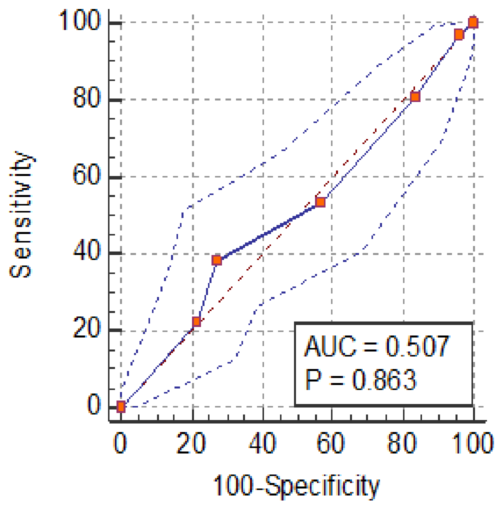

Figure 1B Ordinal composite score Near Faller versus Faller

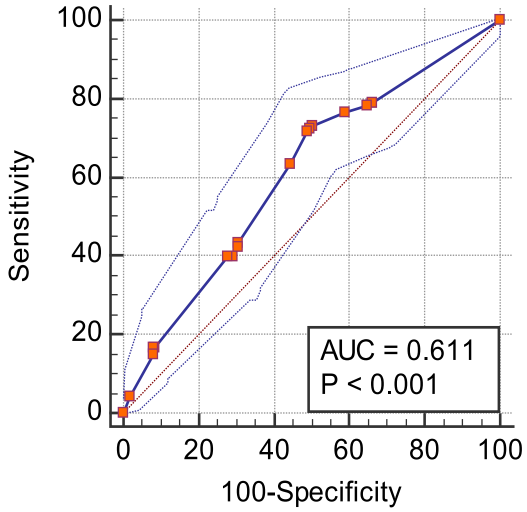

Figure 1C Factor composite score Near Fallers versus Non-Fallers

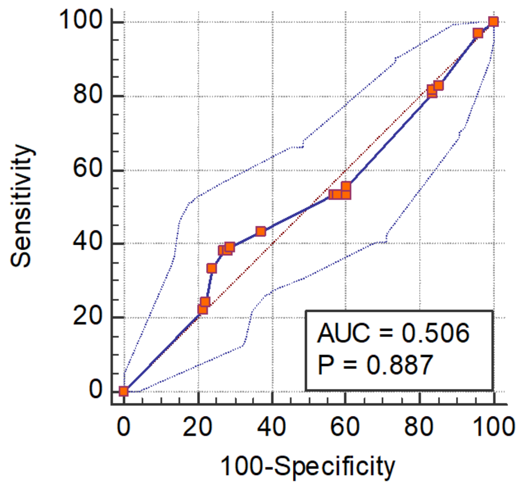

Figure 1D Factor composite score Near Fallers versus Fallers

Table 1. Association, Effect Size, Odds Ratio, Positive and Negative Predictive Values for group status and test outcome, per protocol.

Name of test (Total number of attempts)	Fail/pass attempts			Chi Square (Effect size)	Fall v Non-Fall		Near Fall v Non-Fall		Fall v Near-Fall	
	Non-Fall	Near Fall	Fall	χ^2 (Cramer's V)	Odds Ratio (95%CI)	Positive / Negative Predictive Value (%)	Odds Ratio (95%CI)	Positive / Negative Predictive Value	Odds Ratio (95%CI)	Positive / Negative Predictive Value (%)
SLSEO5s (626)	31/374	21/101	16/83	**12.3** (0.14)**	**2.2** (1.2-4.3)**	34/82	**2.7** (1.5-4.9)**	40/79	0.83 (0.41-1.7)	43/54
Turn 180° (626)	9/396	5/117	7/92	**6.03*** (0.10)	**3.3* (1.2-9.0)**	44/81	2.2 (0.78-6.4)	36/77	1.5 (0.47-4.5)	58/56
Tandem Fwd (627)	15/391	10/112	9/90	**6.79*** (0.10)	**2.5* (1.1-6.0)**	37/81	**2.5* (1.1-5.7)**	40/78	1.0 (0.39-2.5)	47/56
Tandem Bwd (543)	25/343	6/91	6/72	16.35 (0.02)	1.4 (0.63-3.3)	19/83	1.2 (0.50-2.6)	19/79	1.3 (0.45-3.5)	50/56
SLSEO30s (537)	135/230	37/58	29/48	0.13 (0.02)	1.0 (0.62-1.7)	18/83	1.0 (0.66-1.7)	22/80	0.97 (0.52-1.8)	44/55
SLSEC5s (380)	109/146	26/43	27/29	1.41 (0.06)	1.4 (0.78-2.4)	20/83	0.78 (0.45-1.4)	19/77	1.7 (0.86-3.5)	51/60

SLSEC30s (305)	166/35	47/10	41/6	0.63 (0.05)	1.5 (0.57-3.7)	20/85	1.02 (0.47-2.2)	22/78	1.4 (0.47-4.2)	47/63
Hurdle step (624)	21/383	16/106	17/81	**18.6** (0.17)**	**4.0** (2.0-8.0)**	45/83	**2.9** (1.4-5.8)**	43/78	1.4 (0.66-2.9)	52/57
Lunge floor (622)	50/351	32/90	23/76	**16.0** (0.16)**	**2.1** (1.2-3.7)**	32/82	**2.5** (1.5-4.1)**	39/80	0.85 (0.45-1.6)	42/54
Lunge Beam (594)	132/256	59/54	42/51	13.8** **(0.15)**	**1.6* (1.0-2.5)**	45/66	**2.1** (1.4-3.2)**	31/83	0.75 (0.43-1.3)	42/51

Key: EC Eyes Closed; EO Eyes Open; FT Feet Together; Fwd forwards; s seconds; SLS Single Leg Stance; Sn sensitivity; Sp specificity; Tandem heel-touching-toe; *significance ≤ 0.05; **significance ≤ 0.01; Effect size

Appendix 1

Table 1. Association, Effect Size, Odds Ratio, Positive and Negative Predictive Values for group status and test outcome, per protocol.

Name of test	**Fail/pass attempts**			**Chi Square (Effect size)**	**Fall v Non-Fall**		**Near Fall v Non-Fall**		**Fall v Near-Fall**	
(Total number of attempts)	Non-Fall	Near Fall	Fall	χ^2 (Cramer's V)	Odds Ratio (95%CI)	Positive / Negative Predictive Value (%)	Odds Ratio (95%CI)	Positive / Negative Predictive Value (%)	Odds Ratio (95%CI)	Positive / Negative Predictive Value (%)
SLSEO5s (626)	31/374	21/101	16/83	**12.3** (0.14)**	**2.2** (1.2-4.3)**	34/82	**2.7** (1.5-4.9)**	40/79	0.83 (0.41-1.7)	43/54
Turn 180° (626)	9/396	5/117	7/92	**6.03*** (0.10)	**3.3* (1.2-9.0)**	44/81	2.2 (0.78-6.4)	36/77	1.5 (0.47-4.5)	58/56
Tandem Fwd (627)	15/391	10/112	9/90	**6.79*** (0.10)	**2.5* (1.1-6.0)**	37/81	**2.5* (1.1-5.7)**	40/78	1.0 (0.39-2.5)	47/56
Tandem Bwd (543)	25/343	6/91	6/72	16.35 (0.02)	1.4 (0.63-3.3)	19/83	1.2 (0.50-2.6)	19/79	1.3 (0.45-3.5)	50/56
SLSEO30s (537)	135/230	37/58	29/48	0.13 (0.02)	1.0 (0.62-1.7)	18/83	1.0 (0.66-1.7)	22/80	0.97 (0.52-1.8)	44/55
SLSEC5s (380)	109/146	26/43	27/29	1.41 (0.06)	1.4 (0.78-2.4)	20/83	0.78 (0.45-1.4)	19/77	1.7 (0.86-3.5)	51/60
SLSEC30s (305)	166/35	47/10	41/6	0.63 (0.05)	1.5 (0.57-3.7)	20/85	1.02 (0.47-2.2)	22/78	1.4 (0.47-4.2)	47/63
Hurdle step (624)	21/383	16/106	17/81	**18.6** (0.17)**	**4.0** (2.0-8.0)**	45/83	**2.9** (1.4-5.8)**	43/78	1.4 (0.66-2.9)	52/57
Lunge floor (622)	50/351	32/90	23/76	**16.0** (0.16)**	**2.1** (1.2-3.7)**	32/82	**2.5** (1.5-4.1)**	39/80	0.85 (0.45-1.6)	42/54
Lunge Beam (594)	132/256	59/54	42/51	13.8** **(0.15)**	**1.6* (1.0-2.5)**	45/66	**2.1** (1.4-3.2)**	31/83	0.75 (0.43-1.3)	42/51

Key: EC Eyes Closed; EO Eyes Open; FT Feet Together; Fwd forwards; s seconds; SLS Single Leg Stance; Sn sensitivity; Sp specificity; Tandem heel-touching-toe; *significance ≤ 0.05; **significance ≤ 0.01; Effect size

Appendix 2 Intention to treat analysis

Name of test	Association Chi-Square	Effect size	Near Fall v Fall	Near Fall v Non-Fall	Fall v Non-Fall
n = 627	χ^2	Cramer's V	Odds Ratio (95%CI)	Odds Ratio (95%CI)	Odds Ratio (95%CI)
SLS EO 5s	**11.5****	**0.14**	0.9 (0.44-1.83)	**2.53** (1.40-4.60)**	**2.35** (1.22-4.45)**
Turn 180°	5.15	0.09	1.73 (0.53-5.62)	1.70 (0.57-5.08)	**2.94* (1.09-7.93)**
Tandem forward	**6.79***	**0.10**	1.09 (0.42-2.79)	**2.34* (1.02-5.36)**	**2.54* (1.08-5.99)**
Tandem back	**10.7****	**0.13**	1.06 (0.58-1.93)	**1.91** (1.17-3.11)**	**2.02** (1.20-3.39)**
SLS EO 30s	4.36	0.08	0.95 (0.56-1.60)	1.46 (0.97-2.20)	1.38 (0.89-2.14)
SLS EC 5s	1.58	0.05	1.22 (0.69-2.14)	1.06 (0.69-1.62)	1.33 (0.83-2.14)
SLS EC 30s	0.70	0.03	1.08 (0.39-3.01)	4.18 (0.55-2.52)	1.27 (0.55-2.95)
Hurdle step	**18.1****	**0.17**	1.52 (0.74-3.14)	**2.65** (1.35-5.23)**	**4.03** (2.09-7.79)**
Lunge floor	**13.1****	**0.14**	0.83 (0.45-1.53)	**2.39** (1.46-3.90)**	**1.98** (1.16-3.40)**
Lunge Beam	**15.3****	**0.16**	0.73 (0.43-1.25)	**2.18** (1.45-3.29)**	**1.60* (1.03-2.49)**

Key: EC Eyes Closed; EO Eyes Open; FT Feet Together; s seconds; SLS Single Leg Stance; Sn sensitivity; Sp specificity; Tandem heel-touching-toe; *significance p ≤ 0.05; **significance p ≤ 0.01

D TRIPOD Checklist

TRIPOD Checklist: Prediction Model Development

Section/Topic	Item	Checklist Item	Page
Title and abstract			
Title	1	Identify the study as developing and/or validating a multivariable prediction model, the target population, and the outcome to be predicted.	
Abstract	2	Provide a summary of objectives, study design, setting, participants, sample size, predictors, outcome, statistical analysis, results, and conclusions.	
Introduction			
Background and objectives	3a	Explain the medical context (including whether diagnostic or prognostic) and rationale for developing or validating the multivariable prediction model, including references to existing models.	
	3b	Specify the objectives, including whether the study describes the development or validation of the model or both.	
Methods			
Source of data	4a	Describe the study design or source of data (e.g., randomized trial, cohort, or registry data), separately for the development and validation data sets, if applicable.	
	4b	Specify the key study dates, including start of accrual; end of accrual; and, if applicable, end of follow-up.	
Participants	5a	Specify key elements of the study setting (e.g., primary care, secondary care, general population) including number and location of centres.	
	5b	Describe eligibility criteria for participants.	
	5c	Give details of treatments received, if relevant.	
Outcome	6a	Clearly define the outcome that is predicted by the prediction model, including how and when assessed.	
	6b	Report any actions to blind assessment of the outcome to be predicted.	
Predictors	7a	Clearly define all predictors used in developing or validating the multivariable prediction model, including how and when they were measured.	
	7b	Report any actions to blind assessment of predictors for the outcome and other predictors.	
Sample size	8	Explain how the study size was arrived at.	
Missing data	9	Describe how missing data were handled (e.g., complete-case analysis, single imputation, multiple imputation) with details of any imputation method.	
Statistical analysis methods	10a	Describe how predictors were handled in the analyses.	
	10b	Specify type of model, all model-building procedures (including any predictor selection), and method for internal validation.	
	10d	Specify all measures used to assess model performance and, if relevant, to compare multiple models.	
Risk groups	11	Provide details on how risk groups were created, if done.	
Results			
Participants	13a	Describe the flow of participants through the study, including the number of participants with and without the outcome and, if applicable, a summary of the follow-up time. A diagram may be helpful.	
	13b	Describe the characteristics of the participants (basic demographics, clinical features, available predictors), including the number of participants with missing data for predictors and outcome.	
Model development	14a	Specify the number of participants and outcome events in each analysis.	
	14b	If done, report the unadjusted association between each candidate predictor and outcome.	
Model specification	15a	Present the full prediction model to allow predictions for individuals (i.e., all regression coefficients, and model intercept or baseline survival at a given time point).	
	15b	Explain how to the use the prediction model.	
Model performance	16	Report performance measures (with CIs) for the prediction model.	
Discussion			
Limitations	18	Discuss any limitations of the study (such as nonrepresentative sample, few events per predictor, missing data).	
Interpretation	19b	Give an overall interpretation of the results, considering objectives, limitations, and results from similar studies, and other relevant evidence.	
Implications	20	Discuss the potential clinical use of the model and implications for future research.	
Other information			
Supplementary information	21	Provide information about the availability of supplementary resources, such as study protocol, Web calculator, and data sets.	
Funding	22	Give the source of funding and the role of the funders for the present study.	

We recommend using the TRIPOD Checklist in conjunction with the TRIPOD Explanation and Elaboration document.

E Ethics Approval

9 February 2021

HUMAN ETHICS LOW RISK PANEL

APPROVAL NOTICE

Dear Mrs Nicola Baker,

The below proposed project has been **approved** on the basis of the information contained in the application and its attachments.

Project No:	4084
Project Title:	Measuring sway differences between near fallers, fallers and non-fallers
Primary Researcher:	Mrs Nicola Baker
Approval Date:	09/02/2021
Expiry Date:	31/12/2024

Please note: *Due to the current COVID-19 situation, researchers are strongly advised to develop a research design that aligns with the University's COVID-19 research protocol involving human studies. Where possible, avoid face-to-face testing and consider rescheduling face-to-face testing or undertaking alternative distance/online data or interview collection means. For further information, please go to* https://staff.flinders.edu.au/coronavirus-information/research-updates.

RESPONSIBILITIES OF RESEARCHERS AND SUPERVISORS

1. Participant Documentation

Please note that it is the responsibility of researchers and supervisors, in the case of student projects, to ensure that:

- all participant documents are checked for spelling, grammatical, numbering and formatting errors. The Committee does not accept any responsibility for the above mentioned errors.
- the Flinders University logo is included on all participant documentation (e.g., letters of Introduction, information Sheets, consent forms, debriefing information and questionnaires – with the exception of purchased research tools) and the current Flinders University letterhead is included in the header of all letters of introduction. The Flinders University international logo/letterhead should be used and documentation should contain international dialing codes for all telephone and fax numbers listed for all research to be conducted overseas.

2. Annual Progress / Final Reports

In order to comply with the monitoring requirements of the *National Statement on Ethical Conduct in Human Research 2007 (updated 2018)* an annual progress report must be submitted each year on the approval anniversary date for the duration of the ethics approval using the HREC Annual/Final Report Form available online via the ResearchNow Ethics & Biosafety system.

Please note that no data collection can be undertaken after the ethics approval expiry date listed at the top of this notice. If data is collected after expiry, it will not be covered in terms of ethics. It is the responsibility of the researcher to ensure that annual progress reports are submitted on time; and that no data is collected after ethics has expired.

If the project is completed *before* ethics approval has expired please ensure a final report is submitted immediately. If ethics approval for your project expires please either submit (1) a final report; or (2) an extension of time request (using the HREC Modification Form).

For student projects, the Low Risk Panel recommends that current ethics approval is maintained until a student's thesis has been submitted, assessed and finalised. This is to protect the student in the event that reviewers recommend that additional data be collected from participants.

3. Modifications to Project

Modifications to the project must not proceed until approval has been obtained from the Ethics Committee. Such proposed changes / modifications include:

F Study Flyer

Is your *balance* as good as you think?

Flinders UNIVERSITY

inspiring achievement

Have you **tripped** or **slipped** recently?

How **nimble** are you on your feet?

If you are interested in your balance under **fatigued** and **distracted** conditions, are aged between **40-74 years** and have **not fallen** in the past 6 months, come along and have your balance assessed. This is a joint initiative between local government, industry and Flinders University.

Balance assessments will be completed
in Salisbury, Noarlunga, Glenelg and Marion
through **June, July & August 2021**.

For more information please contact Nicky Baker
nicky.baker@flinders.edu.au or 7221 8745.

This project has been approved by the Flinders University Human Research Ethics Committee (Project ID #4084)

inspiring achievement

G Participant Information and Consent Form

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PARTICIPANT INFORMATION SHEET AND CONSENT FORM

Title: 'Measuring sway differences between near fallers, fallers and non-fallers'

Chief Investigator
Mrs Nicky Baker
College of Nursing and Health Sciences
Flinders University
Tel: 7221 8745

Co-Investigator
Professor Sue Gordon
College of Nursing and Health Sciences
Flinders University
Tel: 7221 8746

Co-Investigator
Professor Anthony Maeder
College of Nursing and Health Sciences
Flinders University
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Co-Investigator
Associate Professor Niranjan Bidargaddi
College of Medicine and Public Health
Flinders University
Tel: 7221 8840

Description of the study
This project will investigate balance in middle-aged and young-older aged adults by using a small, wearable inertial sensor on the low back to measure postural sway under different conditions. Postural sway is the movement of the body over the base of support in standing and moving. Wearable inertial sensors are small, lightweight are held in place on the skin by sticky tape and an elastic belt. This project is supported by Flinders University, College of Nursing and Health Sciences.

Purpose of the study
This project aims to

- Identify the relationship between postural sway and near falls
- Measure changes in postural sway as a result of physical fatigue
- Measure changes in postural sway during a concurrent mental task
- Identify the contribution of these sway changes to near falls and falls

inspiring achievement

Measuring sway differences between near fallers, fallers and non-fallers: COVID-safe plan

Potential attendees will be screened prior to testing with the following questions incorporated in the demographic and baseline survey:

1. Covid-19 is still a possibility in our community. Have you potentially been exposed to coronavirus? Yes □ No □
2. Do you have any symptoms such as fever, cough, sore throat, loss of taste/smell or shortness of breath? Yes □ No □

An answer of ‘yes’ precludes attendance. If no, these two questions will be repeated when the participant arrives for testing. An answer of ‘yes’ precludes admission to the testing site.

The participant cohort has been selected to be independently living, without unstable medical or neurological condition, able to mobilise functionally and willing to attend a community venue. The intent is that the participants are relatively healthy and less likely to be a vulnerable group of people.

All testing procedures will follow the current SA Health guidelines for personal and protective equipment which will be checked each day of testing. When the participant arrives at the testing venue, they will be asked the COVID questions and directed to wash/gel their hands. For infection control, the tape that attaches the inertial sensor on the body will be replaced between participants. The inertial sensor and elastic belt that holds it in place will be cleaned with disinfectant wipe before and after contact with each participant. The researcher's hand hygiene will occur per '5 moments' recommended by SA Health (entry to the area, before touching the participant, before placing the sensor and belt on the participant or providing cold drinks, after touching the participant to take off the belt and sensor, and between participants). For the rest and recovery phase of testing, any refreshments such as biscuits will be individually wrapped. Hand hygiene will occur for drink preparation and delivery. The testing area will be wiped down before and after each participant and at the end of the day. This cleaning will include any common touch points such as door handles, light switches etc as well as the general equipment provided by the venue, such as chairs and floor surfaces.

(Wording taken from ethics application and approval Flinders University #4084)

I Survey

SURVEY PAPER COPY: Measuring sway differences between near fallers, fallers and non-fallers.

Unique identifier *First four letters of your street name, two-digit month of birth, last two digits year of birth (E.g. Living in Rockville Avenue, date of birth 1st June 1963 = R O C K 0 6 6 3)*

Demographic information

Date of Birth _ _ /_ _/ _ _ _ _ Age (years) ____ Postcode __ __ __ __

Gender Male □ Female □ Non-binary □

Living status Alone □ With partner □ With others □

Working status Paid work □ Formal voluntary work □ Not in labour force □

Near fall status. A near fall is defined as *any stumble, trip, slip, misstep or other momentary loss of balance where corrective action prevented a fall.*

Have you had any ***near falls*** in the past three months? Yes □ No □

If yes, how many? ____ What happened?

Fear of falling. A fall is defined as *any unexpected event when you lose your balance to come to rest on the ground, floor, or lower level.* Are you afraid of falling? Yes □ No □

Quality of life. In general, I would say my health is:

Excellent Very good Good Fair Poor

Health. Please respond to each question.

I seem to get sick a little easier than other people.

Definitely true Mostly true Don't know Mostly false Definitely false

I am as healthy as anybody I know

Definitely true Mostly true Don't know Mostly false Definitely false

I expect my health to get worse

Definitely true Mostly true Don't know Mostly false Definitely false

My health is excellent

Definitely true Mostly true Don't know Mostly false Definitely false

These questions are about how you feel and how things have been with you for the past 4 weeks. Please give one answer that comes closest to the way you have been feeling. How much time during the past 4 weeks…					
...did you feel full of pep?					
All the time	Most the time	A good bit of the time	Some of the time	A little of the time	None of the time
…did you have a lot of energy?					
All the time	Most the time	A good bit of the time	Some of the time	A little of the time	None of the time
…did you feel worn out?					
All the time	Most the time	A good bit of the time	Some of the time	A little of the time	None of the time
…did you feel tired?					
All the time	Most the time	A good bit of the time	Some of the time	A little of the time	None of the time

The following are about activities you might do during a typical day. Does your health limit you in these activities? If so how much?		
Vigorous activities, such as running, lifting heavy objects, participating in strenuous sports:		
Yes, limited a lot	Yes, limited a little	No, not limited at all
Moderate activities such as moving a table, pushing a vacuum cleaner, bowling, or playing golf:		
Yes, limited a lot	Yes, limited a little	No, not limited at all
Lifting or carrying groceries:		
Yes, limited a lot	Yes, limited a little	No, not limited at all
Climbing several flights of stairs:		
Yes, limited a lot	Yes, limited a little	No, not limited at all
Climbing one flight of stairs:		
Yes, limited a lot	Yes, limited a little	No, not limited at all
Bending kneeling or stooping:		
Yes, limited a lot	Yes, limited a little	No, not limited at all
Walking more than 1.6km (a mile):		
Yes, limited a lot	Yes, limited a little	No, not limited at all

Walking several hundred metres		
Yes, limited a lot	Yes, limited a little	No, not limited at all
Walking 100m (one block):		
Yes, limited a lot	Yes, limited a little	No, not limited at all
Bathing or dressing yourself:		
Yes, limited a lot	Yes, limited a little	No, not limited at all

Hearing

Do you currently wear hearing aids? Yes □ No □

Previous medical or surgical intervention on the ears Yes □ No □

Do you have a history of ear infections Yes □ No □

Exposure to noise at work or in your leisure? Yes □ No □

Do you have tinnitus (ringing or noise in the ears)? Yes □ No □

SSQ5 (Speech, Spatial and Qualities of Hearing)

Please indicate the number that corresponds with your level of agreement for the following questions.

1. When I am having a conversation, I am able to ignore an interfering voice of the same pitch.

(Strongly disagree) 1 2 3 4 5 6 7 8 9 10 (Strongly agree)

2. I am sitting around the table or at a meeting with several people. I can't see everyone. I can tell where any person is sitting as soon as they start speaking.

(Strongly disagree) 1 2 3 4 5 6 7 8 9 10 (Strongly agree)

3. I can tell how far away a bus or truck is, just from the sound.

(Strongly disagree) 1 2 3 4 5 6 7 8 9 10 (Strongly agree)

4. Everyday sounds that I can hear easily, seem clear to me (not blurred).

(Strongly disagree) 1 2 3 4 5 6 7 8 9 10 (Strongly agree)

5. I don't need to concentrate when listening to someone or something.

(Strongly disagree) 1 2 3 4 5 6 7 8 9 10 (Strongly agree)

Dizziness

Do you suffer from dizziness? Yes □ No □ If ***no***, please go straight to vision questions.

If yes, please complete the following questions

Is it constant or does it come and go? Constant □ Comes and goes in spells □

When I'm dizzy, I feel like:

I am spinning in circles Yes □ No □

The world is spinning in circles Yes □ No □

I am nauseated Yes □ No □

My head is swimming Yes □ No □

I am sensitive to light or changes in lighting Yes □ No □

I am sensitive to sounds or changes in sounds Yes □ No □

I lose consciousness Yes □ No □

I had a prolonged headache with light sensitivity and/or nausea Yes □ No □

I had trouble walking in the dark Yes □ No □

Dizziness is in sudden spells with breaks in between Yes □ No □

I get dizzy while sitting or standing still Yes □ No □

I get dizzy when I roll over in bed Yes □ No □

I get dizzy when I turn or move my head Yes □ No □

I get dizzy when I bend over or reach down Yes □ No □

I get dizzy when I stand up quickly from sitting or lying down Yes □ No □

A typical dizzy attack last (how long) _____________ (amount of time duration)

Usually, dizzy attacks happen (how frequently) _______________(amount of time apart)

Vision questions

I consider my vision is Excellent □ Good □ Fair □ Poor □

How good is your eyesight for seeing distance, like a friend across the street? Excellent □ Good □ Fair □ Poor □

Have you had a vision test in the past 12 months? Yes □ No □

Do you normally wear glasses or contact lenses? Yes □ No □

Do you wear bifocal, multifocal or progressive lenses? Yes □ No □

J Instructions for ISWT

"The object of the progressive shuttle walking test is to walk as long as possible, there and back along the 10-metre course, keeping to the speed indicated by the bleeps on the audio recording. You will hear these bleeps at regular intervals.

You should walk at a steady pace, aiming to turn around the cone at one end of the course when you hear the first bleep, and at the other end when you hear the next. At first, your walking speed will be very slow, but you will need to speed up at the end of each minute. Your aim should be to follow the set rhythm for as long as you can.

Each single bleep signals the end of a shuttle, and each triple bleep signals an increase in walking speed. You should stop walking only when you become too breathless to maintain the required speed or can no longer keep up with the set pace.

The test is maximal and progressive. In other words, it is easier at the start and harder at the end. The walking speed for the first minute is very slow. You have 20 seconds to complete each 10-metre shuttle, so don't go too fast. The test will start in 15 seconds, so get ready at the start now. Level one starts with a triple bleep after the 4-second countdown."

To start, the researcher said

> "Walk at a steady pace, aiming to turn around when you hear the signal. You should continue to walk until you feel that you are unable to maintain the required speed. Are you ready? Remember that the object is to walk as long as possible without running"

When the individual was just outside the 0.5m marker they were advised

"Y*ou need to increase your speed to keep up with the test".*

The test was terminated when the participant was more than 0.5m away from the cone when the bleep sounded on a second successive 10m length.

K Borg’s Rate of Perceived Exertion Scale

6

7 Very, very light

8

9 Very light

10

11 Fairly light

12

13 Somewhat hard

14

15 Hard

16

17 Very hard

18

19 Very, very hard

20

L Perceived Recovery Scale

The descriptions of recovery were shown to the participant as follows:

10	Very well recovered / highly energetic
9	
8	Well recovered / somewhat energetic
7	
6	Moderately recovered
5	Adequately recovered
4	Somewhat recovered
3	
2	Not well recovered / somewhat tired
1	
0	Very poorly recovered / extremely tired

M Randomisation

Ppt_num	Bal_Order	Cond_order	DT_Alone	DT_Together
1	SLS,Tandem,Lunge	ISWT - DT	Letter C	328
2	Lunge,SLS,Tandem	ISWT - DT	Letter F	147
3	Tandem,Lunge,SLS	ISWT - DT	Sports	321
4	Tandem,SLS,Lunge	ISWT - DT	119	Animals
5	Lunge,Tandem,SLS	DT - ISWT	366	Transport
6	Lunge,Tandem,SLS	ISWT - DT	170	Body Parts
7	Tandem,SLS,Lunge	DT - ISWT	Clothes	264
8	Lunge,Tandem,SLS	DT - ISWT	Letter P	369
9	Tandem,SLS,Lunge	ISWT - DT	287	Letter R
10	SLS,Lunge,Tandem	ISWT - DT	229	Letter L
11	SLS,Tandem,Lunge	ISWT - DT	360	Gardening
12	Tandem,SLS,Lunge	ISWT - DT	351	Music
13	Lunge,SLS,Tandem	ISWT - DT	Letter W	192
14	SLS,Lunge,Tandem	DT - ISWT	359	Toys
15	Tandem,Lunge,SLS	DT - ISWT	362	Letter G
16	SLS,Lunge,Tandem	ISWT - DT	Letter S	282
17	Lunge,SLS,Tandem	DT - ISWT	333	Letter D
18	Tandem,Lunge,SLS	DT - ISWT	Drinks	182
19	SLS,Tandem,Lunge	DT - ISWT	115	Letter A
20	SLS,Lunge,Tandem	DT - ISWT	124	Letter M
21	Tandem,SLS,Lunge	DT - ISWT	174	Letter B
22	Tandem,SLS,Lunge	DT - ISWT	Colours	173
23	Tandem,Lunge,SLS	DT - ISWT	Letter N	389
24	Lunge,Tandem,SLS	DT - ISWT	Letter T	277
25	Lunge,SLS,Tandem	DT - ISWT	277	Foods
26	Lunge,SLS,Tandem	DT - ISWT	Letter F	144
27	Tandem,Lunge,SLS	ISWT - DT	Letter N	393
28	Tandem,SLS,Lunge	DT - ISWT	Gardening	374
29	Lunge,Tandem,SLS	DT - ISWT	393	Letter L
30	SLS,Tandem,Lunge	ISWT - DT	Animals	349
31	SLS,Lunge,Tandem	ISWT - DT	247	Body Parts
32	Lunge,Tandem,SLS	ISWT - DT	153	Colours
33	Tandem,SLS,Lunge	ISWT - DT	Letter G	213
34	SLS,Lunge,Tandem	ISWT - DT	Letter C	393
35	Tandem,Lunge,SLS	DT - ISWT	Toys	218
36	Tandem,SLS,Lunge	DT - ISWT	189	Letter P
37	Tandem,SLS,Lunge	ISWT - DT	Clothes	383
38	Lunge,Tandem,SLS	ISWT - DT	165	Letter R
39	Tandem,SLS,Lunge	DT - ISWT	Letter A	335

40	Tandem,Lunge,SLS	DT - ISWT	Transport	338
41	Tandem,Lunge,SLS	ISWT - DT	248	Music
42	Tandem,Lunge,SLS	ISWT - DT	Letter S	121
43	SLS,Tandem,Lunge	DT - ISWT	Letter M	131
44	Tandem,Lunge,SLS	ISWT - DT	284	Sports
45	Lunge,Tandem,SLS	DT - ISWT	191	Letter T
46	Lunge,SLS,Tandem	ISWT - DT	396	Foods
47	SLS,Lunge,Tandem	DT - ISWT	347	Letter D
48	Lunge,Tandem,SLS	DT - ISWT	Letter B	214
49	Lunge,Tandem,SLS	DT - ISWT	168	Letter W
50	Tandem,SLS,Lunge	DT - ISWT	Drinks	134
51	Tandem,Lunge,SLS	ISWT - DT	Sports	309
52	Lunge,SLS,Tandem	ISWT - DT	Foods	346
53	Lunge,Tandem,SLS	ISWT - DT	372	Letter N
54	Tandem,Lunge,SLS	DT - ISWT	186	Letter G
55	Tandem,SLS,Lunge	DT - ISWT	283	Letter A
56	Tandem,SLS,Lunge	ISWT - DT	121	Letter P
57	Lunge,SLS,Tandem	ISWT - DT	Clothes	330
58	Lunge,Tandem,SLS	ISWT - DT	290	Music
59	SLS,Lunge,Tandem	DT - ISWT	Animals	156
60	SLS,Lunge,Tandem	DT - ISWT	Letter B	334
61	SLS,Lunge,Tandem	ISWT - DT	Body Parts	292
62	SLS,Lunge,Tandem	ISWT - DT	Letter T	336
63	SLS,Tandem,Lunge	ISWT - DT	Letter F	236
64	Tandem,Lunge,SLS	DT - ISWT	Gardening	294
65	SLS,Lunge,Tandem	DT - ISWT	Transport	353
66	Lunge,SLS,Tandem	DT - ISWT	Letter C	339
67	Lunge,SLS,Tandem	DT - ISWT	Letter M	378
68	SLS,Tandem,Lunge	ISWT - DT	Colours	168
69	SLS,Lunge,Tandem	DT - ISWT	Letter D	168
70	Lunge,Tandem,SLS	ISWT - DT	Drinks	273
71	Tandem,SLS,Lunge	DT - ISWT	Letter L	354
72	Tandem,Lunge,SLS	ISWT - DT	249	Letter W
73	Tandem,Lunge,SLS	DT - ISWT	Toys	395
74	Tandem,Lunge,SLS	ISWT - DT	262	Letter R
75	Lunge,SLS,Tandem	ISWT - DT	280	Letter S
76	Tandem,Lunge,SLS	ISWT - DT	Letter S	169
77	Lunge,Tandem,SLS	ISWT - DT	192	Letter W
78	Tandem,Lunge,SLS	ISWT - DT	351	Drinks
79	Lunge,SLS,Tandem	ISWT - DT	Letter D	252

80	Tandem,Lunge,SLS	DT - ISWT	174	Toys
81	Lunge,SLS,Tandem	DT - ISWT	Letter G	349
82	Lunge,Tandem,SLS	ISWT - DT	Sports	167
83	Tandem,SLS,Lunge	ISWT - DT	269	Animals
84	SLS,Lunge,Tandem	DT - ISWT	249	Letter L
85	SLS,Lunge,Tandem	ISWT - DT	Letter B	174
86	Tandem,Lunge,SLS	ISWT - DT	141	Body Parts
87	Tandem,SLS,Lunge	DT - ISWT	Letter N	387
88	Lunge,SLS,Tandem	DT - ISWT	253	Letter A
89	Tandem,SLS,Lunge	DT - ISWT	306	Letter T
90	Tandem,SLS,Lunge	ISWT - DT	202	Letter C
91	Tandem,SLS,Lunge	DT - ISWT	Clothes	168
92	Lunge,Tandem,SLS	ISWT - DT	Letter R	378
93	SLS,Tandem,Lunge	ISWT - DT	Colours	167
94	SLS,Tandem,Lunge	DT - ISWT	Letter P	107
95	Lunge,Tandem,SLS	DT - ISWT	293	Gardening
96	Lunge,Tandem,SLS	ISWT - DT	Letter F	195
97	SLS,Tandem,Lunge	ISWT - DT	Letter M	145
98	SLS,Lunge,Tandem	ISWT - DT	336	Foods
99	Lunge,SLS,Tandem	DT - ISWT	335	Music
100	SLS,Lunge,Tandem	DT - ISWT	143	Transport
101	Tandem,Lunge,SLS	DT - ISWT	Colours	173
102	Lunge,SLS,Tandem	ISWT - DT	160	Toys
103	Lunge,Tandem,SLS	ISWT - DT	Gardening	224
104	Lunge,Tandem,SLS	ISWT - DT	Body Parts	309
105	Lunge,Tandem,SLS	DT - ISWT	Music	355
106	Tandem,Lunge,SLS	DT - ISWT	Letter F	232
107	Lunge,Tandem,SLS	DT - ISWT	196	Letter G
108	Tandem,SLS,Lunge	DT - ISWT	114	Letter A
109	Lunge,SLS,Tandem	DT - ISWT	Transport	135
110	Tandem,Lunge,SLS	ISWT - DT	Letter P	178
111	SLS,Lunge,Tandem	DT - ISWT	Foods	311
112	SLS,Tandem,Lunge	DT - ISWT	134	Letter L
113	Lunge,Tandem,SLS	ISWT - DT	216	Letter N
114	Tandem,Lunge,SLS	DT - ISWT	366	Animals
115	Tandem,Lunge,SLS	ISWT - DT	256	Letter B
116	SLS,Tandem,Lunge	DT - ISWT	216	Letter R
117	Lunge,Tandem,SLS	ISWT - DT	324	Letter W
118	Tandem,SLS,Lunge	ISWT - DT	191	Letter T
119	Lunge,SLS,Tandem	ISWT - DT	305	Letter C

120	Tandem,Lunge,SLS	ISWT - DT	219	Sports
121	Lunge,SLS,Tandem	DT - ISWT	Letter S	192
122	Tandem,Lunge,SLS	DT - ISWT	320	Drinks
123	SLS,Tandem,Lunge	DT - ISWT	Letter M	229
124	Tandem,Lunge,SLS	ISWT - DT	Letter D	268
125	Tandem,Lunge,SLS	ISWT - DT	326	Clothes
126	Lunge,Tandem,SLS	ISWT - DT	340	Toys
127	Lunge,SLS,Tandem	ISWT - DT	392	Letter T
128	Lunge,SLS,Tandem	DT - ISWT	Letter L	327
129	Lunge,Tandem,SLS	DT - ISWT	Letter B	368
130	Lunge,Tandem,SLS	DT - ISWT	264	Drinks
131	Tandem,SLS,Lunge	DT - ISWT	238	Letter M
132	Lunge,SLS,Tandem	ISWT - DT	Foods	105
133	Tandem,Lunge,SLS	ISWT - DT	Sports	125
134	SLS,Lunge,Tandem	DT - ISWT	224	Letter S
135	SLS,Lunge,Tandem	DT - ISWT	200	Colours
136	Tandem,Lunge,SLS	DT - ISWT	270	Gardening
137	Lunge,Tandem,SLS	ISWT - DT	131	Letter G
138	Lunge,Tandem,SLS	DT - ISWT	Music	338
139	Tandem,SLS,Lunge	ISWT - DT	Letter D	362
140	Lunge,Tandem,SLS	DT - ISWT	Clothes	395
141	Tandem,SLS,Lunge	ISWT - DT	326	Body Parts
142	Tandem,Lunge,SLS	ISWT - DT	Letter W	207
143	SLS,Tandem,Lunge	DT - ISWT	188	Animals
144	Lunge,SLS,Tandem	ISWT - DT	376	Letter R
145	SLS,Lunge,Tandem	DT - ISWT	Letter F	137
146	SLS,Lunge,Tandem	ISWT - DT	Letter N	220
147	Tandem,Lunge,SLS	ISWT - DT	Letter P	370
148	Tandem,SLS,Lunge	ISWT - DT	129	Transport
149	SLS,Lunge,Tandem	ISWT - DT	Letter C	127
150	Lunge,Tandem,SLS	DT - ISWT	Letter A	352
151	Tandem,SLS,Lunge	ISWT - DT	143	Letter C
152	Tandem,SLS,Lunge	ISWT - DT	323	Letter P
153	SLS,Tandem,Lunge	ISWT - DT	Letter T	340
154	Lunge,Tandem,SLS	ISWT - DT	145	Letter F
155	Tandem,SLS,Lunge	ISWT - DT	121	Sports
156	Lunge,Tandem,SLS	ISWT - DT	Letter L	145
157	SLS,Tandem,Lunge	DT - ISWT	226	Letter A
158	SLS,Lunge,Tandem	DT - ISWT	Letter R	364
159	SLS,Tandem,Lunge	DT - ISWT	281	Colours

160	SLS,Tandem,Lunge	ISWT - DT	172	Letter N
161	Lunge,SLS,Tandem	ISWT - DT	Transport	346
162	SLS,Lunge,Tandem	DT - ISWT	Letter B	309
163	Lunge,SLS,Tandem	ISWT - DT	Clothes	238
164	Lunge,SLS,Tandem	ISWT - DT	Drinks	291
165	Tandem,Lunge,SLS	DT - ISWT	Letter G	309

N Data Collection Sheet

Covid-19: Have you potentially been exposed to coronavirus? Yes □ No □
Do you have any symptoms: fever, cough, sore throat, loss taste/smell, shortness of breath? Yes □ No □

If yes to either, please <u>do not</u> enter the premises or participate.

Date	**Venue**				**Time**			
Unique ID	*First four letters of street name*				*Month of birth*		*Year of birth*	

Check consent □ Check survey completed □ Any questions, queries or concerns before starting? □
Coffee in past 2h? □ Vigorous exercise in past 2h? □

Demographics		
Date of birth	Age	
Phone number (txt msg)	Email address	
Height	Weight	Tibial length

Heart Rate	Rest (Sitting 5 min)	Max (208 - (0.7 x age))	Reserve (Max-Rest)	60% Reserve (0.6 x reserve)
Randomization envelope	**Balance order** **1.** **2.** **3.**	**Conditions Order** **1.** **2.** **3. Both ISWT and DT**	**Dual Task Alone** **Dual Task Together**	

Orientation and familiarization			
Sensor	Attached	Comfortable	Working
Balance activities: how to complete, pass/fail	Lunge	SLS	Tandem
Remind participants that they need to stand still with feet together for 5 seconds before, between and after the three balance tests to permit data extraction from the sensor.			
Dual task activities: not scored	Categorical	Subtraction	Prompts
Incremental Shuttle Walk Test	Sound	Speed	Finishing

Thank you for your time. Your participation in this study is very much appreciated.

Test Procedures

Baseline	Balance start time		
Test order	**1**	**2**	**3**
Pass or fail	□	□	□
Lunge score (FMS 3,2,1,0)			

Dual Task Alone (First Second) (circle)	Balance start time		
Which Task?	Comments		
Test order	**1**	**2**	**3**
Pass or fail	□	□	□
Lunge score (FMS 3,2,1,0)			

Incremental Shuttle Walk Test for Fatigue (First Second) (circle)			
ISWT distance (m)	Level 1 □ □ □ Level 2 □ □ □ □ Level 3 □ □ □ □ □ Level 4 □ □ □ □ □ □ Level 5 □ □ □ □ □ □ □ Level 6 □ □ □ □ □ □ □ □ Level 7 □ □ □ □ □ □ □ □ □ Level 8 □ □ □ □ □ □ □ □ □ □ Level 9 □ □ □ □ □ □ □ □ □ □ □ Level 10 □ □ □ □ □ □ □ □ □ □ □ □ Level 11 □ □ □ □ □ □ □ □ □ □ □ □ □ Level 12 □ □ □ □ □ □ □ □ □ □ □ □ □ □ Borg scale RPE Ex HR		
	Balance start time		
Test order	**1**	**2**	**3**
Pass or fail	□	□	□
Lunge score (FMS 3,2,1,0)			

Recovery	1-min	2-min	3-min	4-min	5-min	6-min
Heart rate						
Recovery						

Both ISWT and Dual Task			
ISWT distance (m)	Level 1 □ □ □ Level 2 □ □ □ □ Level 3 □ □ □ □ □ Level 4 □ □ □ □ □ □ Level 5 □ □ □ □ □ □ □ Level 6 □ □ □ □ □ □ □ □ Level 7 □ □ □ □ □ □ □ □ □ Level 8 □ □ □ □ □ □ □ □ □ □ Level 9 □ □ □ □ □ □ □ □ □ □ □ Level 10 □ □ □ □ □ □ □ □ □ □ □ □ Level 11 □ □ □ □ □ □ □ □ □ □ □ □ □ Level 12 □ □ □ □ □ □ □ □ □ □ □ □ □ □ Borg scale RPE Ex HR		
Dual Task:	Balance start time		
Test order	**1**	**2**	**3**
Pass or fail	□	□	□
Lunge score (FMS 3,2,1,0)			

Recovery	1-min	2-min	3-min	4-min	5-min	6-min
Heart rate						
Recovery Scale						

Final balance	start time		
Test order	**1**	**2**	**3**
Pass or fail	□	□	□
Lunge score (FMS 3,2,1,0)			

Home balance exercises required	Yes	No	
Diary	Hard copy	Email	
Text message	Txt	WhatsApp	Messenger
Any final comments?			

O Diary

DIARY **Unique ID**	*First four letters of street name*				*Month of birth*		*Year of birth*	

Please keep a daily entry of any fall or near fall - tick ***only*** if a fall or near fall occurred. When a near fall or fall occurs please provide a brief description. **Fall** *"an unexpected event in which you come to rest on the ground, floor, or lower level"*. **Near fall** *"any stumble, trip, slip, misstep or other momentary loss of balance where corrective action prevented a fall. Include when you land against a wall or a chair instead of the ground"*.

Week 1 (date ----/----/----)	**Monday**	**Tuesday**	**Weds**	**Thursday**	**Friday**	**Saturday**	**Sunday**	Briefly describe what happened (where, how, injury etc)
Fall?								
Near Fall?								
Week 2 (date ----/----/----)	**Monday**	**Tuesday**	**Weds**	**Thursday**	**Friday**	**Saturday**	**Sunday**	
Fall?								
Near Fall?								
Week 3 (date ----/----/----)	**Monday**	**Tuesday**	**Weds**	**Thursday**	**Friday**	**Saturday**	**Sunday**	
Fall?								
Near Fall?								
Week 4 (date ----/----/----)	**Monday**	**Tuesday**	**Weds**	**Thursday**	**Friday**	**Saturday**	**Sunday**	
Fall?								
Near Fall?								
Week 5 (date ----/----/----)	**Monday**	**Tuesday**	**Weds**	**Thursday**	**Friday**	**Saturday**	**Sunday**	
Fall?								
Near Fall?								
Week 6 (date ----/----/----)	**Monday**	**Tuesday**	**Weds**	**Thursday**	**Friday**	**Saturday**	**Sunday**	
Fall?								
Near Fall?								

Week 7 (date ----/----/----)	**Monday**	**Tuesday**	**Weds**	**Thursday**	**Friday**	**Saturday**	**Sunday**	Briefly describe what happened (where, how, injury etc)
Fall?								
Near Fall?								
Week 8 (date ----/----/----)	**Monday**	**Tuesday**	**Weds**	**Thursday**	**Friday**	**Saturday**	**Sunday**	
Fall?								
Near Fall?								
Week 9 (date ----/----/----)	**Monday**	**Tuesday**	**Weds**	**Thursday**	**Friday**	**Saturday**	**Sunday**	
Fall?								
Near Fall?								
Week 10 (date ----/----/----)	**Monday**	**Tuesday**	**Weds**	**Thursday**	**Friday**	**Saturday**	**Sunday**	
Fall?								
Near Fall?								
Week 11 (date ----/----/----)	**Monday**	**Tuesday**	**Weds**	**Thursday**	**Friday**	**Saturday**	**Sunday**	
Fall?								
Near Fall?								
Week 12 (date ----/----/----)	**Monday**	**Tuesday**	**Weds**	**Thursday**	**Friday**	**Saturday**	**Sunday**	
Fall?								
Near Fall?								
Week 13 (date ----/----/----)	**Monday**	**Tuesday**	**Weds**	**Thursday**	**Friday**	**Saturday**	**Sunday**	
Fall?								
Near Fall?								

P Text Messages

Messages will be succinct (less than 160 characters) (Shimoni et al, 2020). They will contain a snippet of evidence-based information about balance and falls prevention to engage participants and minimise annoyance or boredom (Kocielnik & Hsieh, 2017).

1. Stay on your feet with regular physical activity. Have you completed your fall/near fall entries this week?
2. Fall-related injuries caused nearly 25,000 admissions to hospital in SA last year. Don't become a statistic! Please remember the fall/near fall diary.
3. 30% of SA hospital admissions for falls are people aged under 65 years. Keep your balance and keep out of hospital. Please remember the fall/near fall diary.
4. Injuries to hip and knee joints affect balance and walking – weakness and stiffness are barriers to balance. Please complete the fall/near fall diary.
5. Falls from ladders cause serious injury. Keep safe, one step at a time! Please remember the fall/near fall diary.
6. Even modest effort can produce results – keep active to stay on your feet. Have you completed your fall/near fall entries this week?
7. The most common injuries from a fall are hip and thigh fractures, then head injuries. Stay upright and on your feet! Please complete the fall/near fall diary.
8. Look out for your balance and stay on your feet! Have you completed your fall/near fall entries this week?
9. Maintaining good balance is key to staying upright! Have you completed your fall/near fall dairy this week?
10. Physical activity creates healthy minds and bodies – and keeps us on our feet. Have you completed your fall/near fall entries this week?
11. Pets are good for us physically and mentally… but can also be a trip hazard! Please remember to complete the fall/near fall diary entries.
12. How good is your hearing? Did you know good balance relies on good hearing? Please remember to fill in the fall/near diary.

13. A good night's sleep helps us function mentally and physically – and this includes our balance! Please remember to fill in the fall/near diary.
14. Foot pain and footwear choice affect posture, stability, balance and walking. Please remember to complete the fall/near fall diary.

References

Kocielnik, R., & Hsieh, G. (2017). *Send Me a Different Message*. Paper presented at the Proceedings of the 2017 ACM Conference on Computer Supported Cooperative Work and Social Computing.

Shimoni, N., Nippita, S., & Castano, P. M. (2020). Best practices for collecting repeated measures data using text messages. *BMC Med Res Methodol, 20*(1), 2. doi:10.1186/s12874-019-0891-9

Q Floor Surface Analysis

Association between floor surface and balance test results in retrospectively reported near fallers

Test	Lino (pass/fail)	Carpet (pass/fail)	Wood (pass/fail)	Chi-square	Significance p =	Cramer's V Effect size
Base Lun	51/23	40/18	7/1	1.24	0.54	0.09
Base Tan	50/24	48/10	6/2	3.93	0.14	0.17
Base SLS	63/11	52/6	8/0	1.79	0.41	0.11
DT Lun	50/24	44/14	7/1	2.11	0.35	0.12
DT Tan	48/26	45/13	6/2	2.62	0.27	0.14
DT SLS	56/18	50/8	8/0	4.32	0.12	0.18
SW Lun	51/23	45/13	7/1	2.10	0.35	0.12
SW Tan	48/26	42/16	4/4	1.97	0.37	0.12
SW SLS	61/13	49/9	8/0	1.69	0.43	0.11
Both Lun	48/26	43/15	8/0	4.86	0.09	0.19
Both Tan	51/23	36/22	6/2	0.96	0.62	0.08
Both SLS	60/14	49/9	6/2	0.55	0.76	0.06
Final Lun	52/22	42/16	8/0	3.24	1.20	0.15
Final Tan	56/18	47/11	5/3	1.56	0.46	0.11
Final SLS	64/10	53/5	8/0	1.83	0.40	0.11

Key: Base baseline; both distraction with fatigue; DT distraction; Lun lunge; SLS single leg stance; SW (shuttle walk) fatigue; Tan tandem steps